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(71) Applicant (for all designated States except US): GENERAL HOSPITAL CORPORATION [US/US]; 55 Fruit Street, Boston, MA 02114 (US).

(72) Inventors; and

(75) Inventors/Applicants (for US only): TEARNEY, Guillermo, J. [US/US]; 118 Kinnaird Street, #3, Cambridge, MA 02139 (US). YELIN, Dvir [IL/US]; 76 St. Paul Street, Apt. #1, Brookline, MA 02446 (US). BOUMA, Brett, Eugene [US/US]; 12 Monmouth Street, Quincy, MA 02171 (US).

(74) Agent: ABELEV, Gary; Dorsey & Whitney LLP, 250 Park Avenue, New York, NY 10177 (US).

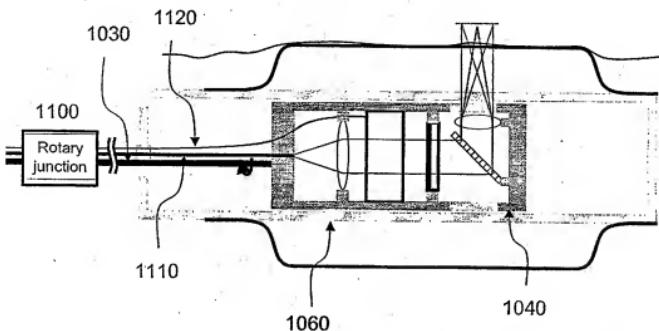
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(54) Title: METHOD AND APPARATUS FOR OPTICAL IMAGING VIA SPECTRAL ENCODING



(57) Abstract: Method, apparatus and arrangement according an exemplary embodiment of the present invention can be provided for generating an image of at least one portion of an anatomical structure. For example, the portion can have an area greater than about 1 mm², and the image can have a resolution a transverse resolution that is below about 10μm. For example, light can be scanned over such portion so as to generate first information which is related to the portion, where the light may be provided through a diffraction arrangement to generate a spectrally dispersed line. Method, apparatus and arrangement according to a further exemplary embodiment of the present invention can be provided for positioning a radiation or optical beam within an anatomical structure based on signals generated by scanning a portion of the structure using the same or a different beam.

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METHOD AND APPARATUS FOR OPTICAL IMAGING VIA SPECTRAL ENCODING

CROSS-REFERENCE TO RELATED APPLICATION(S)

[0001] This application is based upon and claims the benefit of priority from U.S. Patent Application Serial No. 60/721,802, filed September 29, 2005, the entire disclosure of which is incorporated herein by reference.

FIELD OF THE INVENTION

[0002] The present invention relates to devices and methods for comprehensive optical imaging of epithelial organs and other biological structures via spectral encoding.

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BACKGROUND OF THE INVENTION

[0003] Radiological techniques such as X-ray computed tomography ("CT"), magnetic resonance imaging ("MRI"), and ultrasound can enable noninvasive visualization of human pathology at the organ level. Although these modalities may be capable of identifying large-scale pathology, the diagnosis of cancer can require the evaluation of microscopic structures that is beyond the resolution of conventional imaging techniques. Consequently, biopsy and histopathologic examination may be required for diagnosis. Because precancerous growth and early stage cancers often arise on a microscopic scale, they can present significant challenges for identification and diagnosis. Conventional screening and surveillance of these pathologies relies on unguided biopsy and morphological analysis of Hematoxylin and Eosin ("H&E") stained slides. Although this approach may be regarded as a current standard for microscopic diagnosis, it requires the removal of tissue from the patient and significant processing time to generate slides. More importantly, histopathology is inherently a point sampling technique; frequently only a very small fraction of the diseased tissue can be excised and often less than 1% of a biopsy sample may be examined by a pathologist.

[0004] It may be preferable to obtain microscopic diagnoses from an entire organ or biological system in a living human patient. However, the lack of an appropriate imaging

technology can greatly limits options for screening for pre-neoplastic conditions (e.g. metaplasia) and dysplasia. In addition, an inability to identify areas of dysplasia and carcinoma *in situ* has led to screening procedures such as, e.g., random biopsy of the prostate, colon, esophagus, and bladder, etc., which can be highly undesirable and indiscriminate.

5 Many diagnostic tasks presently referred to a frozen section laboratory, such as the delineation of surgical tumor margins, could be improved by a diagnostic modality capable of rapidly imaging large tissue volumes on a microscopic scale. A technology that could fill this gap between pathology and radiology would be of great benefit to patient management and health care.

10 [0005] Technical advances have been made to increase the resolution of non-invasive imaging techniques such as, e.g., micro-CT, micro-PET, and magnetic resonance imaging ("MRI") microscopy. Resolutions approaching 20 μm have been achieved by these technologies, but fundamental physical limitations can still prevent their application in patients. Microscopic optical biopsy techniques, performed *in situ*, have recently been advanced for non-excisional histopathologic diagnosis. Reflectance confocal microscopy ("RCM") may be particularly well-suited for non-invasive microscopy in patients, as it is capable of measuring microscopic structure without tissue contact and does not require the administration of extrinsic contrast agents. RCM can reject out of focus light and detects backscattered photons selectively originating from a single plane within the 15 tissue. RCM can be implemented, e.g., by rapidly scanning a focused beam of electromagnetic radiation in a plane parallel to a tissue surface, yielding transverse or en face images of tissue. The large numerical aperture (NA) that may be used in RCM can yield a very high spatial resolution (1-2 μm), enabling visualization of subcellular structures. High NA imaging, however, can be particularly sensitive to aberrations that 20 arise as light propagates through inhomogeneous tissue. Also, high-resolution imaging with RCM is typically limited to a depth of about 100-400 μm .

25 [0006] RCM has been extensively demonstrated as a viable imaging technique for skin tissue. Development of endoscopic confocal microscopy systems has been more difficult, owing at least in part to the substantial technical challenges involved in miniaturizing a scanning microscope. One major obstacle to direct application of the

concepts of confocal microscopy to endoscopy is the engineering of a mechanism for rapidly rastering a focused beam at the distal end of a small-diameter, flexible probe. A variety of approaches have been proposed to address this problem, including the use of distal micro-electromechanical systems ("MEMS") beam scanning devices and proximal scanning of single-mode fiber bundles. Also, RCM may provide microscopic images only at discrete locations – a "point sampling" technique. As currently implemented, point sampling can be inherent to RCM because it has a limited field of view, which may be comparable to or less than that of an excisional biopsy, and the imaging rate can be too slow for comprehensive large field microscopy.

10 [0007] Another challenge in adapting confocal microscopy to endoscopic applications can include miniaturization of high NA objectives that may be used for optical sectioning. Such miniaturization may be achieved by providing, e.g., a gradient-index lens system, dual-axis objectives, or custom designs of miniature objectives. For example, detailed images of the morphology of cervical epithelium may be obtained in vivo using a fiber optic bundle coupled to a miniature objective lens, and fluorescence-based images of colorectal lesions may be achieved using commercial instruments such as those which may be obtained, e.g., from Olympus Corp. and Pentax/Optiscan.

15 [0008] Despite these advances, there may be a need for improved imaging techniques that can provide microscopic resolution of biological structures *in situ* over large regions.

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OBJECTS AND SUMMARY OF THE INVENTION

[0009] One of the objects of the present invention is to overcome certain deficiencies and shortcomings of the prior art systems and methods (including those described herein above), and provide an exemplary embodiment of a method and an apparatus which are capable of providing comprehensive microscopic optical imaging of anatomical structures such as, e.g., epithelial organs or other bodily tissues.

25 [0010] For example, an apparatus in accordance with exemplary embodiments of the present invention may be in the form of a probe or an assembly, which may be disposable. The probe or assembly may include, e.g., one or more optical waveguides

capable of forwarding an electromagnetic radiation to the probe or assembly and forming an optical beam, one or more focusing arrangements provided at a distal end which may be configured to focus the optical beam, and a scanning arrangement configured to scan the beam across a portion of the anatomical structure. The electromagnetic radiation may 5 include a plurality of wavelengths, and the wavelengths may vary with time. The probe may also include one or more diffraction arrangements which may be configured to diffract or spectrally disperse the beam, one or more correcting arrangements which may be configured to correct for optical aberrations, a mechanism capable of centering or positioning the probe or assembly within the anatomical structure being imaged, and/or a 10 guidewire arrangement which can be capable of translating and/or rotating the probe or assembly. The waveguide may be, e.g., an optical fiber or a bundle of optical fibers or other waveguides. The probe or assembly may further include a spectral encoding arrangement and/or a corrective optical arrangement such as, e.g., a curved transparent surface, which can be used to correct aberrations such as an astigmatism in the optical 15 beam path.

[0011] In certain exemplary embodiments of the present invention, the probe or assembly can be configured to scan a region of the anatomical structure which can have an area greater than about 1 mm², and where the region may include a surface, a volume, or a location below a surface of the anatomical structure. The probe or assembly may be 20 configured to obtain data which can be used to generate an image of the region with a resolution that is below approximately 10μm.

[0012] In further exemplary embodiments of the present invention, a probe or assembly can be provided which is capable of positioning and/or focus the optical beam relative to the anatomical structure. The positioning and/or focusing can be based on, 25 e.g., an interferometric signal, a time-of-flight signal, or an intensity of the electromagnetic radiation. The probe or assembly can include a confocal optical arrangement that can

[0013] In still further exemplary embodiments of the present invention, the probe or assembly can include a locating arrangement that is capable of determining a location of

the probe or assembly relative to a location within the anatomical structure, and an optional positioning arrangement that can control the motion and/or position of the probe based on the location.

5 [0014] In other exemplary embodiments of the present invention, a method for obtaining comprehensive microscopic optical imaging of anatomical structures can be provided, which can include scanning a region of the anatomical structure to be imaged that is larger than about 1 mm² using an electromagnetic radiation such as, e.g., an optical beam, receiving a signal based on the radiation, and generating an image based on the signal, where the image can have a transverse resolution that is below about 10 μm.

10 [0015] In yet further exemplary embodiments of the present invention, a method for positioning or directing an electromagnetic radiation within an anatomical structure is provided, which can include scanning at least a portion of the anatomical structure using the electromagnetic beam, and using a signal which may be based on the electromagnetic radiation to control the position and/or focus of the radiation. A method can also be provided to control the position or focus of a confocal beam within the anatomical structure based on a signal obtained from scanning the electromagnetic radiation over a region of the anatomical structure.

15 [0016] Other features and advantages of the present invention will become apparent upon reading the following detailed description of embodiments of the invention, when taken in conjunction with the appended claims.

BRIEF DESCRIPTION OF THE DRAWINGS

20 [0017] Further objects, features and advantages of the present invention will become apparent from the following detailed description taken in conjunction with the accompanying figures showing illustrative embodiments of the present invention, in which:

25 [0018] FIG. 1 is a schematic illustration of an exemplary spectrally encoded confocal microscopy (SECM) system;

[0019] FIG. 2A is an exemplary SECM image of a swine intestinal epithelium, obtained ex vivo, 100 μm from the tissue surface using a single mode source and single-mode detection (SM-MM) configuration;

5 [0020] FIG. 2B is another exemplary SECM image of a swine intestinal epithelium, obtained using a single-mode source and multi-mode detection (SM-MM) configuration;

[0021] FIG. 2C is a magnified view of an SECM image of a swine intestinal epithelium;

[0022] FIG. 3A is an exemplary SECM image of a swine intestinal epithelium, obtained ex vivo, after compression of the bowel wall at an imaging depth of 50 μm ;

10 [0023] FIG. 3B is an exemplary SECM image of a swine intestinal epithelium, obtained ex vivo, after compression of the bowel wall at an imaging depth of 100 μm ;

[0024] FIG. 4 is a schematic illustration of an exemplary SECM apparatus;

[0025] FIG. 5 is an exemplary SECM image of a USAF chart;

15 [0026] FIG. 6A is an exemplary SECM image based on data taken from a lens paper sample, displayed at a magnification of 1x;

[0027] FIG. 6B is an exemplary SECM image based on data taken from a lens paper sample, displayed at a magnification of 4.5x;

[0028] FIG. 6C is an exemplary SECM image based on data taken from a lens paper sample, displayed at a magnification of 16.7x;

20 [0029] FIG. 6D is an exemplary SECM image based on data taken from a lens paper sample, displayed at a magnification of 50x;

[0030] FIG. 6E is an exemplary SECM image based on data taken from a lens paper sample, displayed at a magnification of 125x;

[0031] FIG. 7 is a series of exemplary SECM data obtained from a lens paper sample at five different focal positions, together with a combine image that was generated by combining the data in the five individual images;

[0032] FIG. 8A is an exemplary SECM image based on data taken from a swine 5 intestinal tissue fragment, displayed at a magnification of 1x;

[0033] FIG. 8B is an exemplary SECM image based on data taken from a swine intestinal tissue fragment, displayed at a magnification of 4x;

[0034] FIG. 8C is an exemplary SECM image based on data taken from a swine intestinal tissue fragment, displayed at a magnification of 20x;

10 [0035] FIG. 8D is an exemplary SECM image based on data taken from a swine intestinal tissue fragment, displayed at a magnification of 40x;

[0036] FIG. 9 is a schematic illustration of an exemplary SECM system capable of imaging large tissue volumes;

15 [0037] FIG. 10 is a schematic illustration of a distal end of an exemplary catheter that may be used for imaging in accordance with exemplary embodiments of the present invention;

[0038] FIG. 11 is a schematic illustration of an exemplary catheter that may be used for imaging in accordance with exemplary embodiments of the present invention that includes an external rotational scanning arrangement;

20 [0039] FIG. 12 A is a schematic illustration of optical effects of a curved window and a negative cylindrical lens;

[0040] FIG. 12B is a schematic illustration of an astigmatic aberration correction using a curved window;

25 [0041] FIG. 13A is an illustration of an exemplary technique which may be used to acquire the desired depth range by stepping through a range of focal depths;

[0042] FIG. 13B is an illustration of an exemplary technique which may be used for imaging tissue at a particular depth by actively adjusting a focal plane;

[0043] FIG. 14A is a schematic illustration of a dual bimorph piezoelectric bender;

[0044] FIG. 14B is a schematic illustration of an exemplary arrangement whereby a 5 motor may be moved within a transparent outer sheath using bending actuators;

[0045] FIG. 15 is a schematic illustration of an exemplary balloon catheter design that is configured to control a focus by translating a collimating lens;

[0046] FIG. 16 is a photograph of a particular variable-focus lens;

[0047] FIG. 17A is a schematic illustration of a cylindrical inner housing design 10 which has a form of a transparent cylinder;

[0048] FIG. 17B is a schematic illustration of a cylindrical inner housing design which includes a transparent window;

[0049] FIG. 17C is a schematic illustration of a cylindrical inner housing design which includes several openings in the housing wall;

15 [0050] FIG. 17D is a schematic illustration of a cylindrical inner housing design which includes openings in a connection between the housing and a motor;

[0051] FIG. 18 is a schematic illustration of electrical and data connections between components of an exemplary imaging system;

20 [0052] FIG. 19A is an illustration of an exemplary probe scanning pattern in which a beam is rotated quickly and simultaneously displaced slowly in an axial direction to provide a spiral imaging pattern;

[0053] FIG. 19B is an illustration of an exemplary probe scanning pattern in which the beam is rotated quickly and then repositioned axially;

[0054] FIG. 19C is an illustration of an exemplary probe scanning pattern in which the beam is rapidly scanned in the axial direction and then repositioned in the rotational direction;

[0055] FIG. 19D is an illustration of an exemplary probe scanning pattern in which the beam is scanned over concentric circular paths cover a circular tissue area;

[0056] FIG. 20A is a schematic illustration of a rapid exchange balloon catheter design which includes a guidewire arrangement located at a distal tip of a housing;

[0057] FIG. 20B is a schematic illustration of a rapid exchange balloon catheter design which includes the guidewire arrangement located at the distal tip of the housing and having a form of a secondary channel;

[0058] FIG. 20C is a schematic illustration of a rapid exchange balloon catheter design which includes the guidewire arrangement located at a proximal tip of a housing and having a form of a secondary channel;

[0059] FIG. 21A is a schematic illustration of a first step in an exemplary technique for positioning a wire balloon catheter that includes insertion of a guidewire;

[0060] FIG. 21B is a schematic illustration of a second step in an exemplary technique for positioning a wire balloon catheter that includes placing a balloon catheter over the guidewire;

[0061] FIG. 21C is a schematic illustration of a third step in an exemplary technique for positioning a wire balloon catheter that includes placing an optical arrangement in the balloon catheter;

[0062] FIG. 22A is a schematic illustration of an exemplary balloon catheter which includes a single channel configured to deliver an inflation material from a remote location to the balloon;

[0063] FIG. 22B is a schematic illustration of an exemplary balloon catheter which includes two sheaths, where the inflation material can be provided between the sheaths;

[0064] FIG. 23A is a schematic illustration of a centering arrangement having a form of a wire cage, where the arrangement is contained within an outer sheath;

[0065] FIG. 23B is a schematic illustration of the centering arrangement having the form of a wire cage, where the arrangement is partially protruding from the outer sheath;

5 [0066] FIG. 23C is a schematic illustration of the centering arrangement having the form of a wire cage, where the arrangement is fully extended from outer sheath;

[0067] FIG. 24A is a schematic illustration of an exemplary SECM/SD-OCT system which includes a wavelength division multiplexer and a dispersion compensator;

10 [0068] FIG. 24B is a schematic illustration of an exemplary spectrum which may be provided by an SECM/SD-OCT system using a linear CCD array;

[0069] FIG. 25 is a schematic illustration of an exemplary SECM/SD-OCT probe;

[0070] FIG. 26 is a schematic illustration of an exemplary SECM/SD-OCT probe which includes a single optical fiber for both the SECM and the SD-OCT arrangements;

15 [0071] FIG. 27 is an exemplary flow diagram of a technique which may be used to adjust a focus for an SECM image using SD-OCT data;

[0072] FIG. 28 is a schematic illustration of a cross section of an exemplary catheter cable;

20 [0073] FIG. 29 is a schematic illustration of an exemplary probe which includes a beam deflection optical arrangement that may provide a more compact probe configuration;

[0074] FIG. 30A is a schematic illustration of a translational scanning technique showing a compact configuration of a probe during delivery of the probe to the site to be imaged;

25 [0075] FIG. 30B is a schematic illustration of the translational scanning technique showing an inner housing of the probe positioned at a distal limit of a translational range;

[0076] FIG. 30C is a schematic illustration of the translational scanning technique showing the inner housing of the probe positioned at a proximal limit of the translational range;

[0077] FIG. 31 is a schematic illustration of an outer housing which includes 5 transparent openings;

[0078] FIG. 32 is a schematic illustration of an exemplary compact probe which includes an off-center collimation optical arrangement and which is configured to provide external rotational scanning;

[0079] FIG. 33A is a schematic illustration of a probe which includes a forward 10 inflatable balloon and an inner housing that is configured to scan while in contact with an inner wall of the balloon;

[0080] FIG. 33B is a schematic illustration of the probe shown in FIG. 33A which is in contact with an inner wall of the inflated balloon;

[0081] FIG. 34A is a schematic illustration of an exemplary probe that includes an 15 outer inflatable balloon and an inner inflatable balloon which may be configured to maintain contact between the probe and a wall of the outer balloon when inflated;

[0082] FIG. 34B is a schematic illustration of the probe shown in FIG. 34A, where the inflated inner balloon is provided around the probe and is configured to maintain contact between the probe and the wall of the inflated outer balloon;

[0083] FIG. 35A is a schematic illustration of a further exemplary probe that includes 20 an outer inflatable balloon and an inner inflatable balloon which may be configured to maintain contact between the probe and a wall of the outer balloon when inflated;

[0084] FIG. 35B is a schematic illustration of the probe shown in FIG. 35A, where 25 the inflated inner balloon is provided between the probe and the outer balloon and is configured to maintain contact between the probe and the wall of the inflated outer balloon;

[0085] FIG. 36A is a schematic illustration of a bottom view of a probe that is configured to scan along a pullback axis while in contact with an inner wall of an inflatable balloon;

[0086] FIG. 36B is a schematic illustration of a side view of the probe shown in FIG. 5 36A;

[0087] FIG. 36C is a schematic illustration of a side view of the probe shown in FIG. 36A, where the probe is in contact with the inner wall of the inflated balloon; and

[0088] FIG. 36D is a front view of the probe shown in FIG. 36C.

[0089] Throughout the figures, the same reference numerals and characters, unless 10 otherwise stated, are used to denote like features, elements, components or portions of the illustrated embodiments. Moreover, while the subject invention will now be described in detail with reference to the figures, it is done so in connection with the illustrative embodiments. It is intended that changes and modifications can be made to the described embodiments without departing from the true scope and spirit of the subject invention as 15 defined by the appended claims.

DETAILED DESCRIPTION OF EXEMPLARY EMBODIMENTS OF INVENTION

[0090] In accordance with exemplary embodiments of the present invention, a method and apparatus for endoscopic confocal microscopy is provided which circumvents the need for miniature, high-speed scanning mechanisms within a probe.

20 Spectrally encoded confocal microscopy ("SECM") is a wavelength-division multiplexed confocal approach that may be used. SECM utilizes a broad bandwidth light source and can encode one dimension of spatial information in the optical spectrum.

[0091] An exemplary SECM technique is shown in FIG. 1. The output from a single-mode optical fiber 100, which may be located at a distal end of a probe, can be collimated 25 by a collimating lens 110, and then illuminate a dispersive optical element (such as, e.g., a transmission diffraction grating 120). An objective lens 130 can then focus each diffracted wavelength to a distinct spatial location within the specimen, resulting in a

transverse line focus 140 where each point on the line may be characterized by a distinct wavelength. After reflection from the specimen, which may be, e.g., biological tissue, the optical signal can be recombined by the diffraction element 120 and collected by the single-mode fiber 100. The core aperture of the single-mode fiber 100 can provide a
5 spatial filtering mechanism that is capable of rejecting out-of-focus light. Outside the probe (and optionally within a system console) the spectrum of the returned light can be measured and converted into confocal reflectance as a function of transverse displacement within the specimen. The spectral decoding can be performed rapidly. Thus an image created by scanning the beam in a direction orthogonal to the line focus
10 can be accomplished by relatively slow and straightforward mechanical actuation.

[0092] SECM techniques may allow the use of endoscopic RCM, and it can be capable of providing image data at extremely high rates using high-speed linear CCD cameras. Commercially available linear CCD arrays can obtain data at a rate greater than about 60 million pixels per second. When incorporated into an SECM spectrometer,
15 these arrays can produce confocal images at speeds that are about 10 times faster than a typical video rate and up to 100 times faster than some endoscopic RCM techniques. The rapid imaging rate and fiber-optic design of typical SECM systems can permit comprehensive, large area microscopy through an endoscopic probe.

[0093] Techniques using optical coherence tomography ("OCT") and variations
20 thereof may be used for comprehensive architectural screening. Acquiring an OCT signal in the wavelength domain, rather than in the time domain, can provide orders of magnitude improvement in imaging speed while maintaining excellent image quality. Using spectral domain OCT ("SD-OCT") techniques, high-resolution ranging can be conducted in biological tissue by detecting spectrally resolved interference between a
25 tissue sample and a reference. Because SD-OCT systems can utilize the same high-speed linear CCD's as SECM systems, they can also be capable of capturing images at 60 million pixels/s, which is approximately two orders of magnitude faster than conventional time-domain OCT ("TD-OCT") systems. With this acquisition rate and resolution, SD-OCT systems can provide comprehensive volumetric microscopy at the architectural
30 level in a clinical environment.

[0094] The information provided by exemplary SD-OCT and SECM systems can be complementary, and a hybrid platform utilizing both techniques can provide information on the architectural and cellular structure of tissue that may be essential to accurate diagnosis. Although a combination of disparate technologies typically requires extensive

5 engineering and may compromise performance, SECM and SD-OCT systems can share key components, and a high-performance multi-modality system can be provided without substantially increasing complexity or cost of the individual systems.

[0095] An SECM system in accordance with certain exemplary embodiments of the present invention can utilize a wavelength-swept 1300 nm source and a single-element

10 photodetector to obtain spectrally encoded information as a function of time. With this system, images can be acquired at rates of up to about 30 frames/second having high lateral (1.4 μm) and axial (6 μm) resolutions, over a 400 μm field of view ("FOV"). Images of freshly excised swine duodenum segments were imaged ex vivo with a high speed system to illustrate the capability of an SECM system to identify subcellular 15 structures that may be found in, e.g., specialized intestinal metaplasia ("SIM") or the metaplastic change of Barretts esophagus.

[0096] FIGS. 2A-2C depict exemplary SECM images of a swine intestinal epithelium obtained ex vivo using two imaging modes and corresponding fiber configurations: a single-mode illumination with single-mode detection ("SM-SM"), and a single-mode

20 illumination with multi-mode detection ("SM-MM"). The SM-SM image in FIG. 2A shows the epithelium structure located 100 μm from the tissue surface using a single mode source and single-mode detection. The image of the same tissue region shown in FIG. 2B, obtained using a single mode source and multi-mode detection (SM-MM) with a core:aperture ratio of 1:4, may have a smoother appearance and may be more 25 easily interpreted because of a reduction in speckle noise. FIG. 2C is a magnified view of the image shown in FIG. 2B that indicates a presence of villi containing a poorly reflecting core (e.g., lamina propria or "lp") and a more highly scattering columnar epithelium. Bright image densities visible at the base of the columnar cells, consistent with nuclei (indicated by arrows) are shown in FIG. 2C.

[0097] The thickness of an esophageal wall being imaged in vivo using OCT techniques can be decreased, e.g., by about a factor of two using an inflated balloon. The swine intestinal sample thickness shown in FIGS. 2A-2C was decreased by the same amount, and the subcellular features observed using SECM techniques were well preserved. FIGS. 3A and 3B show images of this thinned sample obtained at a depth of 50 μ m and 100 μ m, respectively.

[0098] The penetration depth of a commercial 800 nm laser scanning confocal microscope was observed to be reduced by about 20% as compared to that obtained with a 1300 nm SECM system. This reduced penetration may be a result of increased scattering of the shorter wavelength source. Thus an SECM system using an 840 nm source may provide sufficient penetration to identify subcellular structure of, e.g., an intestinal epithelium.

[0099] An apparatus in accordance with certain exemplary embodiments of the present invention that is configured to provide comprehensive SECM images is illustrated schematically in FIG. 4. This exemplary apparatus can be configured to obtain images from a cylindrical sample having a length of 2.5 cm and a diameter of 2.0 cm, which are approximately the dimensions of the distal esophagus. A fiber-coupled 2.0 mW superluminescent diode 200, having a wavelength centered at 800 nm and a bandwidth of 45 nm (QSSL-790-2, qPhotonics, Chesapeake, VA) can be configured to illuminate a 50/50 single-mode fiber optic beam splitter 405. Light transmitted through one port of the splitter can be collimated by a collimator 410 and transmitted through a fiber 412 to a focusing apparatus 415 and to a grating-lens pair that includes a grating 420 (1780 lp/mm, Holographix, LLC, Hudson, MA) and a 350230-B asphere lens 425 (Thor Labs, Inc., Newton, NJ) having a focal length, f , of 4.5 mm, a clear aperture of 5.0 mm, and a NA of 0.55. This arrangement can be capable of producing a 500 μ m longitudinal linear array, or line, of focused, spectrally-encoded spots 430 on an interior surface of the cylindrical sample. The grating-lens pair may be affixed to a shaft of a motor 435 (e.g., a 1516SR, 15 mm diameter motor obtained from MicroMo Electronics, Inc., Clearwater, FL) by a housing 440. As the motor 435 rotates, the spectrally encoded line can be scanned across the inner circumference of the cylindrical sample. The motor 435,

housing 440, and grating-lens pair may be translated along a longitudinal axis of the cylindrical sample during rotation of the motor 435 using, e.g., a computer-controlled linear stage 445 (such as, e.g., a Nanomotion II, 2.5 cm range, obtained from Melles Griot, Rochester, NY). This procedure produced a helical scan of the entire interior 5 surface of the cylindrical sample.

[00100] Light reflected from the sample can be transmitted back through the optical system into the single-mode fiber 412 and provided by the fiber 412 to a spectrometer 450 and linear CCD 455 that can include, e.g., 2048 pixels and has a 30 kHz line rate (such as, e.g., a Basler L104K, obtained from Basler Vision Technologies, Exton, PA). A 10 computer 460 can be used to store, analyze and display image data provided by the spectrometer 450 and CCD 455. Approximately 60,000 points per motor rotation (at 0.5 Hz, or 30 rpm) may be digitized, to provide a circumferential sampling density of approximately 1.0 μ m. The longitudinal velocity of the motor can be approximately 0.25 mm/s and the time required for one complete scan of the cylindrical sample may be about 15 100 seconds.

[00101] The $1/e^2$ diameter of the collimated beam on the grating-lens pair can be about 4.0 mm. As a result, the effective NA of this exemplary apparatus can be approximately 0.4, which corresponds to a theoretical spot diameter of approximately 1.2 μ m and a confocal parameter of approximately 2.5 μ m. In a system that is free of optical 20 aberrations, a theoretical spectral resolution on the sample may be 0.8 \AA , which can yield up to approximately 630 resolvable points across the spectrally encoded line 430. The spectrometer 450 in the detection arm can be designed to exceed the predicted spectral resolution of the probe.

[00102] An SECM scan of a 1951 USAF resolution chart obtained using this apparatus 25 is shown in FIG. 5. The smallest bars in this Figure, which are separated by 2.2 μ m, were resolved. A transverse line spread function full-width-half-maximum ("FWHM") and an axial FWHM function obtained using a mirror scanned through the focus were measured as 2.1 μ m and 5.5 μ m, respectively. The field of view was observed to be about 500 μ m. These measurements were slightly lower than corresponding theoretical values, which

may be attributed to aberrations in the optical path. These parameters indicate that the exemplary apparatus described herein can be capable of providing sufficient resolution to be used for confocal microscopy in biological tissue.

[00103] Exemplary SECM image data for a complete pullback image of a 2.5 cm 5 phantom specimen are shown in FIG. 6. Polar coordinates were converted to rectangular coordinates prior to generating these displayed images. The phantom specimen was made using lens paper affixed to the inner surface of a 2.1 cm inner diameter Teflon tube. In a low magnification image shown in FIG. 6A, macroscopic structure of the paper, including folds and voids, can be observed. Circumferential stripes that are visible may 10 have resulted from the lower spectral power and lens aberrations that may be present at or near the ends of the spectrally-encoded line. Individual fibers and fiber microstructure can be clearly resolved in regions of this data set that are presented at higher magnifications, as shown in FIGS. 6B-6E.

[00104] By adjusting the focusing apparatus 415 in FIG. 4A, cylindrical two-dimensional ("2D") images of the phantom sample were acquired at five discrete focal 15 depths over a range of 120 μm . These five images 710-750 shown in FIG. 7 were then summed to create an integrated image 760, which demonstrates a nearly complete coverage of the surface of the phantom sample.

[00105] Imaging biological samples using an SECM apparatus such as that described 20 herein can be complicated by the lack of a centering apparatus for the optical scan head. In order to provide further improvements for generating wide-field microscopy images and data, a sample of swine intestine was placed on top of a 2.0 cm diameter transparent cylinder. A 360° scan of this sample, which was acquired in 1 second, is shown in FIG. 25 8A. Imaged tissue likely appears in only one sector of the cylindrical scan because the probe was not centered and the sample did not wrap completely around the cylinder. FIGS. 8B-8D show a sequence of exemplary magnified regions of this tissue sample. The image shown in FIG. 8B is an expansion of a 1.5 cm sector outlined by a dotted rectangle in FIG. 8A. Similarly, the image in FIG. 8C represents an expansion of the rectangle outlined in FIG. 8B, and the image in FIG. 8D represents an expansion of the

rectangle outlined in FIG. 8C. Magnified images of the tissue in the image FIG. 8B are suggestive of a glandular structure. The magnified images in FIGS. 8C-8D exhibit villi and nuclear features that are similar to those observed using a 1300 nm SECM system, as shown in FIGS. 2 and 3. Other areas of the SECM scan in FIG. 8A show artifacts, 5 including specular reflectance from the transparent cylinder and complete signal dropout, both of which may result from improper positioning of a focused SECM beam.

[00106] Conducting comprehensive confocal microscopy in patients can present a variety of technical challenges. Such challenges may include, e.g., increasing the imaging rate, miniaturizing the probe optical components and mechanical components, 10 incorporating a centering mechanism, and implementing a technique for dynamically changing the focal plane.

[00107] The image acquisition speed of an SECM system can be improved by, e.g., a factor of about 2-4 as compared with the exemplary system described hereinabove. Such an improvement can be realized by providing certain modifications. For example, a 15 higher power semiconductor light source (such as, e.g., a Superlum Diode, T-840 HP: 25 mW, 840 nm, 100 nm spectral bandwidth) can provide, e.g., approximately 1000 spectrally resolvable points. Such an increase in optical power can improve sensitivity and a larger bandwidth may widen the field of view, making it possible to scan the SECM beam approximately two times faster. Also, using an optical circulator such as, 20 e.g., an OC-3-850 (Optics for Research, Caldwell, NJ) can increase the efficiency of light delivered to the probe and collected from the probe. Using a faster, more sensitive linear CCD such as, for example, an AVIIVA M4-2048 having 2048 pixels and a 60 kHz readout rate (Atmel Corporation,) can provide a twofold increase in data acquisition speed and an improved spectral response over the wavelength range used to generate 25 image data. Performance may also be improved by using, e.g., a Camera Link interface that can be capable of transferring data at a rate of approximately 120 MB/s from a camera to a hard-drive array for storage.

[00108] Sensitivity, which can be understood to refer to a minimum detectable reflectance, is a system parameter that can affect confocal image quality and penetration

depth. A fraction of the incident light, which may be approximately 10^{-4} to 10^{-7} , can be reflected from skin at depths up to approximately 300 μm when using a near-infrared RCM technique. Based on the NA of the objective lens used in the exemplary system in accordance with certain exemplary embodiments of the present invention described herein, and the observation that skin may attenuate light more significantly than non-keratinized epithelial mucosa, the exemplary SECM probe objective described herein may collect approximately 3×10^{-4} to 3×10^{-7} of the illuminating light reflected from deep within tissue. A 25 mW light source may be separated into, e.g., approximately 1000 independent beams. A maximum double pass insertion loss can be estimated to be approximately 10 dB (which can include a 6 dB loss from the probe, and a 4 dB loss from the fiber optics and spectrometer). Each pixel in an array may thus be illuminated by approximately 50 to 50,000 photons/pixel for each line integration period based on these estimated parameters.

[00109] Using a multi-mode detection technique, a factor of 10 signal gain may be achieved, resulting in approximately 500 to 500,000 photons/pixel per scan for such a configuration. A single pixel on an Atmel AV11VA M4 camera, e.g., can reliably detect light if a signal is above the dark current fluctuation that occurs at approximately 240 photons. If this device has approximately a 50% quantum efficiency at these wavelengths, a minimum detectable signal can be produced at approximately 480 photons/pixel per scan. Based on these approximations, an Atmel camera may have sufficient sensitivity to allow SECM imaging at deeper tissue depths. Quantum noise-limited detection of a predicted minimum reflectance can be achieved by using a multi-mode fiber for collection or by increasing the source power.

[00110] A schematic diagram of an apparatus capable of performing large-area microscopic imaging of epithelial organs in accordance with certain exemplary embodiments of the present invention is shown in FIG. 9. A light source 900, which may be a broadband source or a wavelength swept source, can provide light which may be conveyed through a circulator 910 or, alternatively, through a fiber splitter. The light can then be transmitted to an imaging catheter 930 through a scanning mechanism 920. Scanning can be performed either externally to the catheter or within the catheter. In

certain preferred exemplary embodiments, pullback scanning may be performed outside the catheter, and rotational scanning may be performed inside the catheter. Reflected light that is collected may then be detected with a detector 940 which may be, e.g., a spectrometer if a broadband light is used. The detector 940 may also be, e.g., a single detector if a wavelength swept source is used. Data provided by the detector 940 may be processed, displayed and/or saved by a computer 950 which may also be configured to control and synchronize the scanning procedure.

[00111] Screening large luminal organs may preferably utilize a centering of a distal portion of a catheter within the lumen to provide a consistent focus distance and/or depth 10 relative to the tissue, and rapid acquisition of circumferential images over lengths of several centimeters. These criteria can be satisfied by incorporating a circumferentially scanning imaging probe within a centering device. Provided an imaging optical arrangement located at or near the middle of the centering device can provide several additional advantages, including, e.g., elimination of surface height fluctuations, which 15 may simplify focusing requirements, and physical coupling of the imaging system to a patient, which can greatly reduce motion artifacts that may otherwise occur.

[00112] A schematic diagram of the distal end of an SECM catheter in accordance with certain exemplary embodiments of the present invention is shown in FIG. 10. Light can be provided through an optical fiber 1000, which may be fixed by a fiber chuck 1005, 20 and then collimated using a collimating lens 1010. This light may then pass through a variable focusing mechanism 1015 and a cylindrical lens 1020 that can be configured to pre-compensate the optical path to correct for astigmatism effects. The light may then be diffracted through a diffraction grating 1025, which can be configured to diffract a center wavelength of the light by, for example, approximately 90 degrees, and focused by an 25 imaging lens 1030 onto a spectral encoded line 1035.

[00113] Speckle artifact may be reduced using multi-mode detection by increasing the diameter of a pinhole aperture associated with the optical fiber 1000. This technique can provide an increased signal throughput and a reduction in speckle artifacts, together with only a slight decrease in spatial resolution. A double clad optical fiber may be used to

implement this technique for spectral encoding, in which a single-mode core can illuminate a tissue and a multi-mode inner cladding can detect reflected light.

[00114] The imaging lens 1030 may preferably have a relatively large working distance that can be, e.g., approximately 2-7 mm, and maintain a large NA of 5 approximately 0.25 to 0.5. In addition, the imaging lens 1030 can be thin, preferably not more than about 5 mm thick. Conventional lenses, such as aspheres or achromats, may be used as imaging lenses.

[00115] The inner housing 1040 may surround some or all of the various optical components and the motor 1045, and it may allow for longitudinal positioning of these 10 components within the outer housing 1060. The inner housing 1040 can include portions thereof that have good optical transmission characteristics and low wavefront distortion to allow high quality imaging, while still maintaining structural rigidity to maintain a motor shaft 1050 centered within the probe. Materials that may be used to form transparent windows as part or all of the inner housing 1040 may include, for example, 15 glass or plastic materials such as, e.g., Pebax and high-density polyethylene (HDPE).

[00116] The outer housing 1060 can surround the inner housing 1040, and can be configured to remain in a fixed position relative to the imaged tissue 1080 using the centering mechanism 1065. An opening in a wall of the outer housing 1060 can allow a pullback cable 1065 to move the inner housing 1040. Linear scanning can be conducted 20 by affixing the inner housing 1040 to a computer-controlled translator (such as a translator that may be provided, e.g., by Newport Corp., Irvine, CA), while maintaining the outer housing 1060 in a fixed position relative to the tissue 1080 being imaged. Such a pullback technique may be used, e.g., to obtain longitudinal esophageal OCT images. All or a portion of the outer housing 1060 may be transparent to allow a transmission of 25 light therethrough. Optical characteristics of the transparent portions of the outer housing 1060 can be similar to those of the inner optical window 1055.

[00117] The cylindrical lens 1020, the diffraction grating 1025, and the imaging lens 1030 may be housed in a rotational housing 1070, which may be attached to the motor shaft 1050. A conventional motor 1045 may be used, which can have a diameter as small

as about 1.5 mm or less. Using an encoder may improve image quality and registration, and may also increase the diameter of the motor 1045 to approximately 6-10 mm. Such a motor can be provided, e.g., by (MicroMo Electronics, Inc. (Clearwater, FL). Dimensions of motor wires can be minimized to limit obstruction of a field of view of the

5 apparatus. Circumferential scanning may be performed by rotating the rotational housing 1070 within the inner housing 1040 using the motor 1045 via the motor shaft 1050.

[00118] A catheter configured to provide rotation of the inner housing 1040 relative to the external housing 1060 from a location external to a distal end of the catheter, in accordance with certain exemplary embodiments of the present invention, is illustrated

10 schematically in FIG. 11. A rotary motion can be transmitted through an optical rotary junction 1100, and light may be coupled into a rotation optical fiber 1110. The rotary junction may also maintain electrical contact via one or more electrical wires 1120 and mechanical contacts via a rotatable pullback cable 1030 that can be configured to control pullback and focusing mechanisms. In the exemplary apparatus configuration shown in
15 FIG. 11, the inner housing 1140 does not surround a motor and thus it can be smaller and lighter.

[00119] A cylindrical lens may be used to correct for astigmatism effects that can be created by a wall of a balloon or another centering device and/or by a transparent window or a transparent section of the inner and/or outer housing. A curved glass can induce

20 astigmatism in a manner similar to that of a negative cylindrical lens. For example, the astigmatism induced by the two curved transparent walls shown in FIG. 12A are optically similar to the negative cylindrical lens shown towards the right side of this Figure. Light passing through the central dashed line of any of the objects shown in FIG. 12A may have a shorter path than light passing through the upper or lower dashed lines, which
25 leads to induced astigmatism. Efficient and accurate correction of this optical distortion can be achieved, e.g., by placing a curved window, similar to the window that induces the astigmatism, in the optical path, as shown in FIG. 12B. The curvature axis of the correcting curved window should be perpendicular to the axis of the curved housing windows to provide optical correction of the astigmatism.

[00120] In another exemplary embodiment of the present invention, an endoscopic SECM system can be provided that is capable of comprehensively imaging an organ without user intervention during the acquisition of image data. The system can be capable of accounting for motion due to, e.g., heartbeat, respiration, and/or peristalsis movements. Utilization of a centering mechanism can greatly reduce artifacts caused by motion of the tissue being imaged. For example, variations in distance between an imaging arrangement and the tissue being imaged can vary, for example, by as much as approximately $\pm 250 \mu\text{m}$ during one comprehensive scan. This distance variation can occur on a slow time scale (e.g., over several seconds) relative to a circumferential scanning speed, but it may be significant relative to a time required to scan the length of a tissue region being imaged during longitudinal pullback of the imaging arrangement.

[00121] An exemplary technique can be used in accordance with certain exemplary embodiments of the present invention to reduce or eliminate the effects of tissue motion during sampling. This technique, illustrated in FIG. 13A, can include a procedure for obtaining image data over a wider range of focal depths. If a desired total imaging depth is, for example, $200 \mu\text{m}$, and a variation in tissue distance from the imaging arrangement is, e.g., $\pm 250 \mu\text{m}$, then image data can be acquired over a focal range of about $700 \mu\text{m}$. This procedure can ensure that image data is obtained throughout the desired tissue volume. Although many portions of the volumetric image may not contain tissue when imaged, it is likely that at least one good image would be obtained from most regions of the tissue volume of interest.

[00122] A second exemplary technique that may be used to compensate for motion of tissue during imaging is illustrated in Fig. 13B. This technique can include a procedure for determining a distance between the imaging lens and a surface of the tissue being imaged. This distance can be tracked, and a focus of the lens can be adaptively controlled to provide a known focal distance relative to the tissue surface throughout the acquisition of image data in the tissue volume of interest. Adaptive focusing can decrease the number of focal scans required, and therefore may also decrease the time needed to obtain comprehensive coverage of the tissue volume of interest. Focus of the

beam can be controlled, e.g., using an interferometric signal, a time-of-flight signal, an intensity of the electromagnetic radiation, etc.

[00123] The above-described exemplary techniques for addressing motion of the tissue being imaged can utilize a mechanism for adjusting the focal distance of the imaging

5 arrangement. There are several exemplary techniques that may be used for adjusting the focal depth within the tissue volume being imaged. For example, an inner housing of the imaging arrangement that includes a focus lens can be moved relative to an exterior housing. To achieve this motion, for example, multi-layered bimorph piezoelectric actuators 1410 (e.g., D220-A4-103YB, Piezo Systems, Inc., Cambridge, MA) shown in

10 FIG. 14A can be attached to, e.g., a metal sheet 1420 at both ends, which may provide a buckling of the ceramic material. These actuators can be placed back-to-back, as shown in FIG. 14A, which can effectively double the range of their free motion. Four such actuators 1430 can be arranged between an outer sheath 1440 and an assembly 1450 that can include a motor and focal optical components surround the motor, as shown in FIG.

15 14B. These actuators 1430 can be utilized to change the focal position over the required range by controllably displacing the assembly 1450 relative to the outer housing 1440. This technique can require the presence of a high voltage within the probe, additional electrical wires that may traverse and interrupt the field of view, and/or an increase of the overall diameter of a probe containing the imaging arrangement by, e.g., several mm.

20 [00124] An alternate exemplary technique that may be used to adjust the focal distance of the imaging arrangement is shown in FIG. 15. A cable housing 1510 can be provided that surrounds a cable 1530. The cable 1530 can be attached at one end to a collimating lens 1540, which may be configured to be movable in a longitudinal direction relative to a housing 1550. The collimating lens 1540 can be moved relative to the housing 1550

25 and other optical components to vary the focal distance. This translation can be controlled, e.g., externally to the imaging catheter, using the cable 1530 as is illustrated in FIG. 15. Alternatively, motion of the collimating lens 1540 can be controlled, e.g., by an electric or piezoelectric motor that can be provided inside the catheter. The focal distance can also be varied by moving an optical fiber 1520, which can provide the light used to image tissue, relative to the collimating lens 1540. Alternatively, both the optical

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fiber 1520 and the collimating lens 1540 may be moved relative to each other to vary the focal distance.

[00125] The focal length can be shifted by a distance Δz by changing the separation between the optical fiber 1520 and the collimating lens 1540 by a distance of 5 approximately $M^2 \Delta z$, where M is a magnification factor of the imaging apparatus. For example, an exemplary imaging apparatus can have a magnification factor that is approximately 3. To obtain a change in the focal distance of approximately $\pm 450 \mu\text{m}$, the distance between the optical fiber 1520 and the collimating lens 1540 would need to move approximately $\pm 4.0 \text{ mm}$, which is a distance that can be achieved using any of the 10 techniques described above for changing the focal distance.

[00126] A further exemplary technique that can be used to vary the focal distance can be to utilize an electronically tunable variable lens. For example, a commercially available lens 1600 (Varioptic AMS-1000, Lyon, France) shown in FIG. 16, which may be used in cell phone cameras, may be utilized to vary the focal length in an imaging 15 apparatus in accordance with an exemplary embodiment of the present invention. This lens 1600 uses an electrowetting principle, and can provide a variable focal length between about -200 mm and 40 mm, with optical quality that may only be limited by diffraction effects. The current effective clear aperture (CA) of this exemplary lens 1600 is 3.0 mm and the total outer diameter (OD) is 10 mm. A similar lens having a 4.0 mm 20 CA and a 6.0 mm OD may be possible to produce. The full-range response time of this exemplary lens 1600 is about 150 ms, which can be sufficiently fast to be used to track the distance between the optical components and the tissue surface and adjust the focal distance accordingly. It may be possible to produce this type of lens having a response time of about 10 ms. Utilizing a variable lens such as the one described above between 25 the collimator and the SECM grating can provide, e.g., a focal distance that can vary by about $\pm 300 \mu\text{m}$ or greater.

[00127] Various configurations can be provided for the inner housing in accordance with certain exemplary embodiments of the present invention. For example, a housing formed from transparent material 1700 can be used, as shown in FIG. 17A.

Alternatively, a housing can be provided that includes a transparent window 1710, as shown in FIG. 17B. A housing may also be provided that includes an opening 1720 between two walls, such as that as shown in FIG. 17C, or an opening adjacent to a motor 1730 that may be attached to the housing as shown, e.g., in FIG. 17D.

5 [00128] An exemplary schematic diagram of a control and data recording arrangement which can be used with the exemplary system shown in FIG. 9 is provided in FIG. 18. The arrangement shown in FIG. 18 can be configured to record a beam position while acquiring imaging data 1800, which can provide a more precise spatial registration of the imaging data 1800. As shown in FIG. 18, the imaging data 1800 can be acquired by a

10 data acquisition and control unit 1810. A catheter scanner arrangement may scan a beam, e.g., using a rotary motor 1820 to provide angular motion of the beam and a pullback motor 1830 to move the beam longitudinally. The rotary motor 1820 can be controlled by a rotary motor controller 1840, and the pullback motor 1830 can be controlled by a pullback motor controller 1850. Each of these control techniques may be performed

15 using a closed loop operation. The data acquisition and control unit 1810 can direct the motor controller units 1840, 1850 to provide specified motor velocities and/or positions. Encoder signals generated by the motors 1820, 1830 can be provided to both the motor controller units 1840, 1850 and the data acquisition and control unit 1810. In this manner, the encoder signals associated with each motor 1820, 1830 can be recorded

20 when a line of imaging data 1800 is acquired, thereby allowing a precise beam position to be associated with each line of data 1800.

[00129] Various scanning priorities that may be used in the imaging catheter in accordance with an exemplary embodiment of the present invention are shown in FIG. 19. For example, an exemplary scanning technique in which rotational scanning is

25 performed as a first priority and axial (pullback) scanning is performed as a second priority is shown in FIG. 19A. This technique can provide a set of data having a helical geometry. In a further scanning technique, the axial scanning can be performed in small increments, with each axial increment following a full revolution, as shown in FIG. 19B. Alternatively, axial (pullback) scanning can be performed as a first priority and rotational scanning can be performed as a second priority, which may generate the scanning pattern

shown in FIG. 19C. A greater imaging quality can be achieved along a direction of the first scan priority. Thus, a choice of scan priority may depend on whether transverse (rotational) images or axial images are preferred. Imaging of other organs or tissues that may have different symmetries can be performed in several ways. For example, a 5 circular scanning pattern that may be used to image certain organs is shown in FIG. 19D.

[00130] In a further exemplary embodiment of the present invention, a balloon catheter such as, e.g., the one shown in FIG. 10, can be configured to allow for a rapid-exchange placement procedure using a guidewire. In a rapid-exchange placement procedure, a guidewire can first be placed in an organ to be imaged, and the catheter can 10 then be threaded down the guidewire. This procedure can allow easier and more precise placement of the catheter in many applications. Various configurations may be used to guide a catheter using a rapid-exchange procedure. For example, FIG. 20A shows an exemplary guidewire 2000 that passes through a hole 2010 in a distal end of the outer housing 2040. In a second exemplary configuration shown in FIG. 20B, a guidewire 2000 15 passes through a tube 2020 that is attached to the distal end of the outer housing 2040. Alternatively, the guidewire 2000 can be configured to pass through the tube 2020 which may be attached to a proximal end of the outer housing 2040, as shown in FIG. 20C.

[00131] An exemplary procedure that may be used to position a catheter that employs a guidewire in a center lumen of the catheter is illustrated in FIGS. 21A-C. First, the 20 guidewire 2100 can be placed within the organ 2150, as shown in FIG. 21A. Next, an outer housing 2110 of the catheter, together with a balloon 2120, can be threaded over the guidewire 2100, as shown in FIG. 21B. Finally, the inner housing 2130, which may contain an optical arrangement, can be threaded down the catheter center lumen as shown in FIG. 21C, and an imaging procedure using the optical arrangement can be performed.

25 [00132] Two exemplary configurations of a balloon catheter are shown in FIG. 22. In FIG. 22A, a device 2200 that may include a source of pressurized air or gas can be used to inflate a balloon 2210. A tube or other small passageway 2230 can be provided that is connected to the balloon 2210 surrounding the catheter and which allows transfer of the pressurized air or gas to the balloon 2210. Pressure within the balloon 2210 being

inflated can be monitored using a manometer 2220. This pressure can be used to optimize the balloon inflation as well as to assess placement of the catheter by monitoring pressure within a surrounding organ which may be contacted by the inflated balloon 2210. Alternatively, a passageway 2240 can be provided along an outer sheath of the catheter, which can allow transfer of the pressurized air or gas to the balloon 2210, as shown in FIG. 22B. A balloon that is capable of changing its diameter in response to pressure changes may be used, where focus depth can be controlled by varying the balloon diameter and thus moving the surrounding tissue to be allows transfer of the pressurized air or gas to the balloon 2210, with respect to the imaging lens.

10 [00133] An exemplary catheter design that may be used in accordance with another exemplary embodiment of the present invention is shown in FIGS. 23A-23C. This catheter design can be configured to use one or more expandable wire strands 2300 to center an inner optical core of an imaging device within a luminal organ. The catheter may include an additional sheath 2310 and a set of expandable wire strands 2300 located 15 within the sheath 2310 that may be provided around the outer housing 2320, as shown in FIG. 23A. After placement of the catheter, the wire strands 2300 can be pushed through the sheath 2310 to protrude from the end thereof as shown in FIG. 23B. Alternatively, the sheath 2310 can be retracted from the outer housing 2320. A sufficient length of the wire strands 2300 can be exposed around the outer housing 2320 to allow the wire strands 2300 to expand the surrounding organ or tissue as shown in FIG. 23C, and to center the housing 2320. After the imaging procedure is performed, the wire strands 2300 may be 20 pulled back into the sheath 2310 and the catheter can be removed.

25 [00134] Exemplary OCT and RCM techniques can reject or ignore multiply scattered light received from a tissue sample being imaged, and thereby detect singly backscattered photons that may contain structural information. Each of these techniques, however, can reject multiply scattered light in a different way.

[00135] For example, the RCM techniques may employ confocal selection of light reflected by tissue being imaged from a tightly focused incident beam. RCM techniques can be implemented by rapidly scanning the focused beam in a plane parallel to the tissue

surface, which may provide transverse or en face images of the tissue. A large numerical aperture (NA), which can be used with conventional RCM techniques, may yield a very high spatial resolution (e.g., approximately 1-2 μm that can allow visualization of subcellular structure. Imaging procedures using a high NA, however, can be particularly 5 sensitive to aberrations that can arise as light propagates through inhomogeneous tissue. Therefore, high-resolution imaging using RCM techniques may be limited to a depth of about 100-400 μm .

[00136] The OCT techniques can utilize coherence gating principles for optical sectioning and may not rely on the use of a high NA lens. OCT techniques may thus be 10 performed using an imaging lens having a relatively large confocal parameter. This can provide a greater penetration depth into the tissue being imaged (e.g., approximately 1-3 mm) and a cross-sectional image format. These advantages may come at the expense of a reduced transverse resolution, which can be typically on the order of about 10-30 μm .

[00137] Thus, in view of the distinctions described above, the exemplary OCT and 15 RCM techniques can offer different imaging information which may be complementary. For example, RCM techniques can provide subcellular detail, whereas OCT techniques can provide, e.g., architectural morphology. Imaging information from these two size regimes can be critical for histopathologic diagnosis, and in many cases, it may be difficult if not impossible to make an accurate diagnosis without using both. Although a 20 combination of these disparate imaging techniques may conventionally utilize extensive engineering efforts which can compromise performance, SECM and SD-OCT techniques can share certain components. Therefore, a high-performance multi-modality system employing both of these imaging techniques can be provided that does not include a substantial increase in complexity or cost relative to a system that may use either 25 technique alone.

[00138] An overview of an exemplary system that is capable of performing both SECM techniques and SD-OCT techniques in accordance with an exemplary embodiment of the present invention is shown in FIG. 24A. In this exemplary system, a portion of a broadband light source bandwidth can be used for obtaining SECM image

data, and a further portion of the bandwidth data can be used, e.g., to obtain SD-OCT data. For example, a light source 2400 can be used to provide electromagnetic energy having a bandwidth greater than, e.g., about 100 nm. Devices that may be used as a light source 2400 can include, e.g., a diode-pumped ultrafast laser (such as that available from, e.g., IntegralOCT, Femtolasers Produktions GmbH, Vienna, Germany), or an array of super luminescent diodes (which may be obtained, e.g., from Superlum, Russia).

[00139] A portion of the light source spectrum that may be used for SD-OCT data (e.g., light having a wavelength between about 810-900 nm) can be separated from a portion of the spectrum that may be used for SECM data using a wavelength division

10 multiplexer (WDM) 2410 and transmitted to a catheter 2420 and to a reference arm 2445. Light returning from the catheter 2420 through an SECM optical fiber 2430 and an SD-OCT optical fiber 2440 can be provided to a spectrometer 2450. The spectrometer 2450 may be configured so that approximately half of the elements of the exemplary CCD array 2460 shown in FIG. 24B can detect a signal associated with the SECM data, and 15 approximately half of the CCD elements can detect a signal associated with the SD-OCT data. The SD-OCT data can be converted into axial structural data, e.g., by performing a Fourier transformation following interpolation of the SD-OCT data from wavelength space to k-space. For example, if the spectrometer 2450 has a resolution of approximately 0.1 nm, a total SD-OCT ranging depth may be greater than about 2.0 mm.

20 Axial image resolution using the SD-OCT technique may be approximately 5 μ m.

[00140] A schematic overview of an exemplary SECM/SD-OCT probe is shown in FIG. 25. This probe is similar to the probe shown, e.g., in FIG. 15, and it further includes an arrangement configured to provide an SD-OCT beam path. In order to obtain an SD-

25 OCT beam, an OCT optical fiber 2500 can be inserted into the inner housing, together with an SECM optical fiber 2510. The OCT optical fiber 2500 can be configured to illuminate a small lens 2520. A confocal parameter and a spot size for the SD-OCT beam can be selected to achieve cross-sectional imaging over a range of depths. Exemplary values of the confocal parameter spot size can be, e.g., be approximately 1.1 mm and 25 μ m, respectively. The NA of the SD-OCT lens 2520 can be selected to be, e.g., approximately 0.02, and a collimated beam diameter of the SD-OCT beam can be

selected to be, e.g., approximately 200 μm . A dichroic mirror 2530 can be placed before the SECM grating to reflect the SD-OCT light beam 2540 and transmit the SECM light beam 2550. The dichroic mirror 2530 shown in FIG. 25 is arranged at an angle of approximately 45 degrees with respect to the SD-OCT light beam 2540. This angle can

5 be increased by using an appropriate coating on the mirror 2530, which can allow the SD-OCT beam 2540 to overlap the SECM beam 2550 for a more precise spatial registration of the two images. Optical aberrations of the SD-OCT beam 2540 which may be produced, e.g., by a curved window or balloon can be corrected by using a cylindrical element to pre-compensate for astigmatism as shown in FIG. 12B.

10 [00141] A further exemplary embodiment of a catheter probe which may be used for both SECM imaging and SD-OCT imaging is shown in FIG. 26. Broadband light may be provided through a single optical fiber 2600, instead of through two separate fibers 2500, 2510 as shown in FIG. 25. A portion of the light which may be used to form an SD-OCT beam 2640 may be reflected out of the optical path of the SECM beam 2650 using a

15 dichroic mirror 2610. The diameter of the SD-OCT beam 2640 may be reduced by an aperture 2620 and/or by focusing the SD-OCT beam 2640 using a lens 2630. The SD-OCT arrangement may also be used to locate a surface of a tissue being imaged using an SECM technique, even with SD-OCT depth resolutions between about 20-100 μm . This can be performed even if the bandwidth of the SD-OCT beam 2640 is not sufficient to

20 obtain a high quality SD-OCT image.

[00142] Data obtained from an exemplary SD-OCT image can be used to adjust a focal plane of an SECM beam. An exemplary flow diagram illustrating this technique is shown in FIG. 27. For example, SD-OCT image data may be obtained from a depth scan (step 2700) and subsequently processed (step 2710). The image data may be analyzed and displayed as an SD-OCT image (step 2720). This image data may also be used to determine the location of a tissue surface (step 2730) using, for example, edge detection algorithms. Once the surface location of the tissue has been determined, a variable focus mechanism can be used to adjust a location of a focal plane of the SECM arrangement (step 2740). This focus control technique can be performed rapidly (e.g., in less than about 100 ms), which may allow for real-time tracking and focusing of a tissue surface.

A location of a tissue edge can be calibrated using an angle that is formed with respect to the SECM beam.

[00143] A cross section of an exemplary catheter cable 2800 which may be used with certain exemplary embodiments of the present invention is shown in FIG. 28. The cable

5 2800 may include, e.g., a pullback cable 2810, a plurality of wires 2820 configured to supply electric power to a motor, a focus control cable 2830, a channel 2840 configured to provide a gas or other fluid to an inflatable balloon or membrane, an SECM optical fiber 2850, and/or an SD-OCT optical fiber 2860.

[00144] A schematic illustration of an exemplary SECM probe 2900 is shown in FIG.

10 29. The probe 2900 includes two prisms 2910 which may be configured to deflect a beam 2920 before it passes through a grating 2930 and an imaging lens 2940. This exemplary configuration can provide more space within the probe 2900 for the objective lens 2940, which can result in a higher NA and/or a size reduction of the probe 2900.

[00145] A further reduction in probe length can be achieved using the exemplary

15 probe configuration 3000 shown in FIGS. 30A-30C. The probe 3000 can include an inner housing 3010 which may be provided within an outer housing 3020 while the probe 3000 is delivered to the imaging location, as shown in FIG. 3A. After the probe 3000 is placed and centered within the tissue or organ to be imaged, the inner housing 3010 can slide through the outer housing 3020 to provide an extended pullback range, as shown in
20 FIGS. 30B and 30C. For example, providing an imaging lens 3020 near a center of the inner housing 3010 can provide increased positional stability at the extreme scanning locations shown in FIGS. 30B and 30C.

[00146] An exemplary outer housing 3100 is shown in FIG. 31. The outer housing 3100 can be made of rigid materials such as, e.g., stainless steel or plastic. It may include

25 one or more gaps 3110 which can allow light to pass therethrough to generate image data without introducing optical aberrations. Optionally, the gaps 3110 may include transparent windows.

[00147] FIG. 32 shows an exemplary probe in accordance with certain exemplary embodiments of the present invention. The probe 3200 can provide a compact configuration of components and a small overall probe size. For example, a cylindrical inner housing 3210 can be configured to rotate and move freely within a cylindrical outer housing 3220, allowing a collimating lens 3230 and an optical fiber 3240 to be placed away from a center axis of the inner housing 3210. Scanning of a region of tissue to be imaged can be performed externally, where motion of the inner housing 3210 can be controlled using a pullback cable 3250.

[00148] In certain exemplary embodiments of the present invention, a liquid such as, e.g., water or an index-matching oil can be provided in a space between an imaging lens and a surface of the tissue to be imaged. Providing such a liquid can, e.g., improve optical parameters such as a NA and/or reduce back reflections of a light beam used to obtain image data.

[00149] An exemplary probe configuration 3300 which can provide a high NA for obtaining image data is shown in FIGS. 33A and 33B. For example, an inner housing 3310 can be provided in an outer housing 3320, which may also include an uninflated balloon 3330. The uninflated balloon 3330 may be inflated such that it can expand forward of the outer housing 3320. The inner housing 3310 may then be deployed outside of the outer housing 3310 and within the inflated balloon 3340. An elastic arrangement 3350 can be provided in a compressed configuration between the inner housing 3310 and the outer housing 3320, as shown in FIG. 33A. The elastic arrangement 3350 can be configured to position the inner housing 3310 against an inside wall of the inflated balloon 3340 when the inner housing 3310 is deployed, as shown in FIG. 33B. The inner housing 3310 can be configured to scan a region of tissue outside of the inflated balloon 3340 the balloon area using a pullback cable 3360. The cable 3360 can be capable of controlling both rotation and longitudinal translation (e.g., pullback) of the inner housing 3310 within the inflated balloon 3340. Spacers 3370 may be used to improve contact between the imaging optical arrangement and the wall of the inflated balloon 3340 or the adjacent tissue surface.

[00150] A further exemplary probe configuration 3400 is shown in FIGS. 34A and 34B, which can be capable of maintaining an inner probe housing 3410 against an inside wall of an outer balloon 3420, in accordance with certain exemplary embodiments of the present invention. For example, an outer balloon 3420 and an inner balloon 3430, shown 5 uninflated in FIG. 34A, can be provided such that they surround the inner housing 3410. Each balloon may be inflated, as shown in FIG. 34B. In this exemplary configuration, the inner housing 3410 may be attached to one face of the inner balloon 3430. Rotational and translational scanning within the outer balloon 3420 may be performed by moving the inner housing 3410 together with the inner balloon 3430 relative to the outer balloon 10 3420.

[00151] A still further exemplary probe configuration 3500 is shown in FIGS. 35A and 35B, which can be capable of maintaining an inner probe housing 3510 against an inside wall of an outer balloon 3520, in accordance with certain exemplary embodiments of the present invention. The outer balloon 3520, shown uninflated in FIG. 35A, may be 15 inflated within an organ or region of tissue to be imaged. An inner balloon 3530, shown uninflated in FIG. 35A, may be provided between the inner housing 3510 and the outer balloon 3520. The inner balloon 3530 may be inflated, as shown in FIG. 35B, and pressure provided by the inner balloon 3530 can be used to maintain contact between the inner housing 3510 and an inner wall of the outer balloon 3520, as shown in FIG. 35B. 20 The exemplary probe configurations 3400 and 3500 shown in FIGS. 34 and 35, respectively, may be used without an outer housing. The uninflated balloons 3420, 3430, 3520, 3530 may be packed inside an external enclosure that can be used to deliver the probe 3400, 3500 to a desired location. Such an external enclosure can optionally be formed, e.g., from a dissolvable material.

[00152] An exemplary configuration of an SECM probe 3600 is shown in FIGS. 36A-36D, which is capable of providing a spectrally encoded line 3610 that lies perpendicular to an axis of an organ or a balloon cylinder. A bottom view of this probe configuration is provided in FIG. 36A, and a corresponding side view is shown in FIG. 36B. FIG. 36C shows a further side view in which the probe housing 3640 is deployed within an inflated 25 balloon 3650, similar to that shown in FIG. 33B. In this exemplary configuration, a

longitudinal (e.g., pullback) direction can be a primary scanning direction, such that the probe housing 3640 is moved in this longitudinal direction at a relatively fast rate of speed. Scanning in a rotational direction around a longitudinal axis can be performed at a relatively low rate compared to the longitudinal speed. The probe 3600 can be provided 5 with positioning arrangements such as those shown, e.g., in any of FIGS. 33-35. The probe housing 3640 can include a mirror 3620 which may be configured to deflect a light beam towards a suitably positioned grating to provide a spectrally-encoded line 3610 configured as shown in FIGS. 36A and 36D.

[00153] Combination of SD-OCT and SECM imaging arrangements within a probe 10 can provide a useful apparatus for obtaining structural information on different scales using different image formats. Data obtained for both imaging techniques can be acquired simultaneously, because the resolutions of the two techniques are different. However, useful scan rates for the two techniques may not be compatible with each other. For example, a typical SECM scan rate can be provided using a rotation rate, e.g., of 15 about 1 Hz and a longitudinal pullback speed, e.g., of approximately 1 mm/s. Typical scan rates for obtaining SD-OCT image data can be, e.g., approximately 50-100 Hz in a rotational direction and, e.g., approximately 0.2 - 0.5 mm/s in a longitudinal direction.

[00154] One technique which may be used to obtain comprehensive image data that is 20 properly sampled for both techniques is to conduct an additional comprehensive SD-OCT scan, sampled appropriately, following acquisition of the SECM data set. This technique may increase the data acquisition time for a tissue region by, e.g., approximately 1-2 minutes. Encoder signals obtained for both the rotating and linearly translating motors can be digitized throughout each scan. The encoder signals can be corrected for shifts in position of a balloon by quantitatively correlating SD-OCT images to determine angular 25 and rotational offsets for each scan. This technique can provide accurate spatial registration of the SD-OCT and SECM data sets within about 500 μ m.

[00155] In a further exemplary embodiment of the present invention, an imaging arrangement provided, e.g., in a probe may be operated in an abbreviated imaging mode (e.g., 'scout imaging') to determine if a catheter which may be used to deliver the probe

is properly positioned within the organ or tissue region to be imaged. A comprehensive set of image data can be obtained after proper catheter placement is confirmed.

[00156] In a still further exemplary embodiment of the present invention, a balloon centering catheter may be inflated using a material that is optically transparent other than

5 air such as, e.g., water, heavy water (D2O), oil, etc. A lubricating agent may also be used to aid insertion of the catheter. In certain exemplary embodiments of the present invention, a mucousal removal agent may be applied prior to obtaining image data to reduce the amount of mucous present in the organ to be imaged, where presence of such mucous may reduce image quality.

10 [00157] The foregoing merely illustrates the principles of the invention. Various modifications and alterations to the described embodiments will be apparent to those skilled in the art in view of the teachings herein. Indeed, the arrangements, systems and methods according to the exemplary embodiments of the present invention can be used with any OCT system, OFDI system, SD-OCT system or other imaging systems, and for
15 example with those described in International Patent Application PCT/US2004/029148, filed September 8, 2004, U.S. Patent Application No. 11/266,779, filed November 2, 2005, and U.S. Patent Application No. 10/501,276, filed July 9, 2004, the disclosures of which are incorporated by reference herein in their entireties. It will thus be appreciated that those skilled in the art will be able to devise numerous systems, arrangements and
20 methods which, although not explicitly shown or described herein, embody the principles of the invention and are thus within the spirit and scope of the present invention. In addition, to the extent that the prior art knowledge has not been explicitly incorporated by reference herein above, it is explicitly being incorporated herein in its entirety. All publications referenced herein above are incorporated herein by reference in their
25 entireties.

What Is Claimed Is:

1. An apparatus comprising:
 - 5 at least one first arrangement configured to forward electromagnetic radiation to an anatomical structure, and continuously scan an entire region of at least one portion of the anatomical structure using the electromagnetic radiation to generate at least one particular signal, wherein the entire region has an area that is greater than 1 mm²; and
 - 10 at least one second arrangement configured to receive the particular signal and to generate at least one image based on the particular signal that has a transverse resolution that is below 10µm.
2. The apparatus according to claim 1, wherein the entire region is a volume of the structure.
- 15 3. The apparatus according to claim 1, wherein the at least one image is contiguous without substantial gaps therein.
4. The apparatus according to claim 1, wherein the at least one portion is provided
 - 20 on a surface of the anatomical structure.
5. The apparatus according to claim 1, wherein the at least one portion is provided below a surface of the anatomical structure.
- 25 6. The apparatus according to claim 1, wherein the electromagnetic radiation comprises a plurality of wavelengths.
7. The apparatus according to claim 1, wherein the electromagnetic radiation comprises one or more wavelengths that vary over time.

8. The apparatus according to claim 1, wherein at least one of the at least one first arrangement or the at least one second arrangement includes a microscope arrangement, and where the microscope arrangement is at least one of a multiphoton microscope arrangement or a confocal microscope arrangement.

5

9. The apparatus according to claim 8, wherein the microscope arrangement includes a spectral encoding arrangement.

10. The apparatus according to claim 1, wherein the at least one anatomical structure is an internal organ.

11. The apparatus according to claim 1, wherein at least one of the at least one first arrangement forwards the electromagnetic radiation via an optical fiber arrangement, or at least one second arrangement receives the particular signal via an optical fiber arrangement.

15

12. The apparatus according to claim 11, wherein the optical fiber arrangement includes a plurality of electromagnetic radiation guiding arrangements.

20 13. The apparatus according to claim 1, wherein the particular signal is associated with at least one portion of an intensity of the electromagnetic radiation received from the anatomical structure.

25 14. The apparatus according to claim 1, wherein the at least one first arrangement is provided in a probe.

15. The apparatus according to claim 1, wherein the first arrangement includes at least one optical component which is configured to compensate for at least one optical aberration.

30

16. The apparatus according to claim 15, wherein the at least one optical component

comprises a curved surface.

17. The apparatus according to claim 16, wherein the at least one optical aberration is an astigmatism.

5

18. The apparatus of claim 1, further comprising a positioning arrangement configured to position at least one of the at least one first arrangement or the at least one second arrangement at a particular location relative to the anatomical structure.

10 19. The apparatus according to claim 1, wherein the at least one first arrangement is further configured to position a focus of the electromagnetic radiation at a plurality of depths within the anatomical structure.

20. The apparatus according to claim 1, further comprising at least one third

15 arrangement configured to:

generate a further signal;

determine at least one location of a particular section associated with the anatomical structure based on the further signal; and

20 control at least one of a motion or a position of a focus of the at least one electromagnetic radiation to a further location within the anatomical structure based on the further signal.

21. The apparatus according to claim 20, wherein the further signal is the particular signal.

25

22. The apparatus according to claim 21, wherein the further signal is at least one of an interferometric signal, a time-of-flight signal, or an intensity of the electromagnetic radiation.

23. The apparatus according to claim 20, wherein the further signal is associated with at least one distance between at least one location within the region of the anatomical structure and at least one component of the at least one first arrangement.

5 24. An apparatus comprising:

at least one first arrangement configured to forward at least one electromagnetic radiation to an anatomical structure, and scan at least one portion of the anatomical structure using the at least one electromagnetic radiation to generate at least one signal; at least one second confocal arrangement; and

10 at least one third arrangement configured to automatically control a position of a focus of the at least one second confocal arrangement as a function of the at least one signal to a predetermined location within the anatomical structure.

15 25. The apparatus according to claim 24, wherein the at least one signal is at least one of an interferometric signal, a time-of-flight signal associated with the at least one electromagnetic radiation, or an intensity of the at least one electromagnetic radiation.

26. The apparatus according to claim 24, further comprising:

20 at least one third arrangement configured to receive the at least one signal and a further signal from a reference to generate the interferometric signal; and at least one fourth arrangement which is capable of generating the at least one electromagnetic radiation, where the electromagnetic radiation comprises a plurality of wavelengths.

25 27. The apparatus according to claim 24, wherein the at least one first arrangement is at least one of a confocal microscope arrangement or a multi-photon microscope arrangement.

30 28. The apparatus according to claim 24, wherein the at least one first arrangement includes a spectral encoding arrangement.

29. The apparatus according to claim 24, wherein the at least one anatomical structure is an internal organ.
30. The apparatus according to claim 24, wherein at least one of (i) the at least one first arrangement forwards the electromagnetic radiation via an optical fiber arrangement, or (ii) the at least one second arrangement receives the particular signal via an optical fiber arrangement.
31. The apparatus according to claim 30, wherein the optical fiber arrangement includes a plurality of electromagnetic radiation guiding arrangements.
32. The apparatus according to claim 24, wherein the at least one electromagnetic radiation comprises a plurality of wavelengths.
- 15 33. The apparatus according to claim 24, wherein the at least one electromagnetic radiation comprises one or more wavelengths that vary over time.
34. An apparatus comprising:
 - at least one first arrangement configured to forward at least one electromagnetic radiation to an anatomical structure, and to scan at least one portion of the anatomical structure using the electromagnetic radiation to generate data; and
 - 20 at least one second arrangement configured to:
 - a) generate a signal based on a determination of at least one location of a particular section of the anatomical structure; and
 - b) control at least one of a motion or a position of a focus of the at least one electromagnetic radiation relative to the at least one location within the anatomical structure as a function of the signal.
- 25 35. The apparatus according to claim 34, wherein the particular section is provided on a surface of the anatomical structure.

36. The apparatus according to claim 34, wherein operations (a) and (b) are performed substantially simultaneously.

37. The apparatus according to claim 34, wherein the at least one second arrangement 5 is configured to position the focus of the at least one electromagnetic radiation at a plurality of depths within the anatomical structure.

38. The apparatus according to claim 34, wherein the at least one second arrangement controls a depth of the focus within the anatomical structure.

10

39. The apparatus according to claim 34, wherein the at least one location is determined based on an interferometric signal.

15

40. The apparatus according to claim 39, wherein the interferometric signal is based at least in part on the data.

41. A method for imaging a region of an anatomical structure, comprising:

forwarding at least one electromagnetic radiation to an anatomical structure;

20

continuously scanning an entire region of at least one portion of the anatomical structure using the electromagnetic radiation, wherein the entire region has an area that is greater than 1 mm²;

obtaining at least one particular signal based on the electromagnetic radiation; and

25

generating at least one image based on the particular signal, wherein the image has a transverse resolution that is below 10μm.

42. A method for positioning an electromagnetic radiation within an anatomical structure, comprising:

30

forwarding at least one electromagnetic radiation to an anatomical structure;

scanning at least one portion of the anatomical structure using the at least one electromagnetic radiation;

obtaining at least one particular signal based on the electromagnetic radiation;

5 controlling a position of a focus of the at least one electromagnetic radiation based on the at least one particular signal to a predetermined location within the anatomical structure.

43. A method for positioning an electromagnetic radiation within an anatomical 10 structure, comprising:

forwarding at least one electromagnetic radiation to an anatomical structure;

scanning at least one portion of the anatomical structure using the at least one electromagnetic radiation;

15 obtaining data based on the at least one electromagnetic radiation; generating a signal based on a determination of at least one location of a particular section of the anatomical structure; and

controlling at least one of a motion or a position of a focus of the at least one electromagnetic radiation relative to the at least one location based on the signal.

20

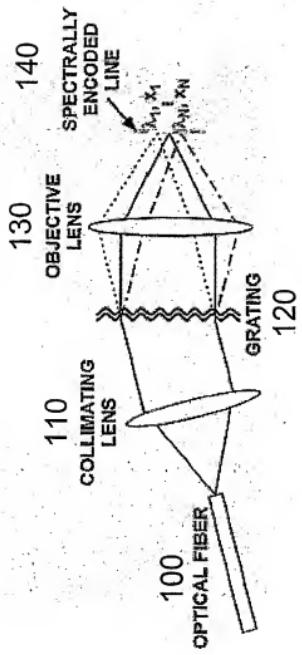


FIG. 1



FIG. 2A

FIG. 2B

FIG. 2C

Figure 2 consists of three panels of fluorescence microscopy images. Panel A shows a dense, granular pattern with a 100 μm scale bar. Panel B shows a more diffuse, textured pattern with a 100 μm scale bar. Panel C shows a pattern with two white arrows pointing to specific spots, labeled 'lp', with a 50 μm scale bar.

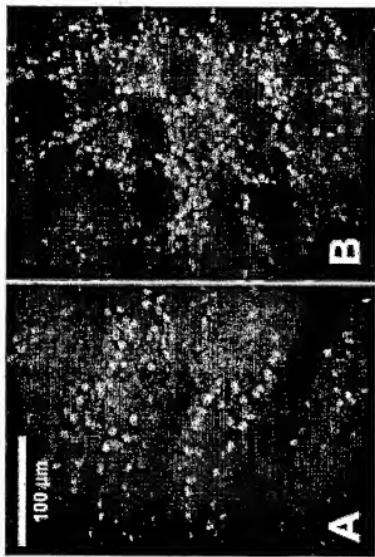


FIG. 3A
FIG. 3B

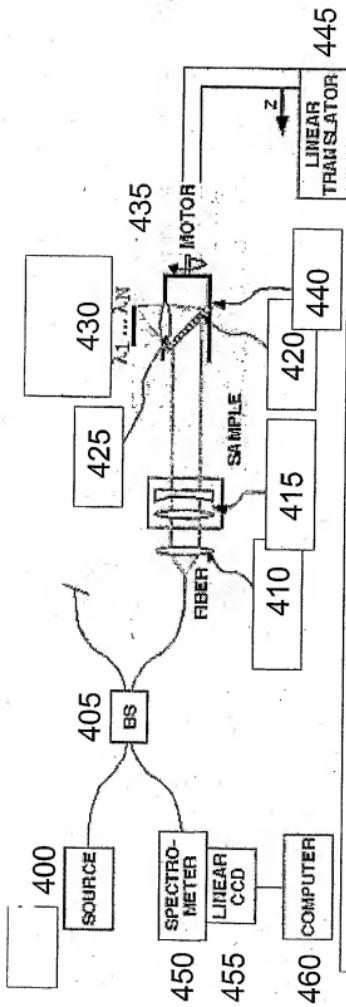


FIG. 4

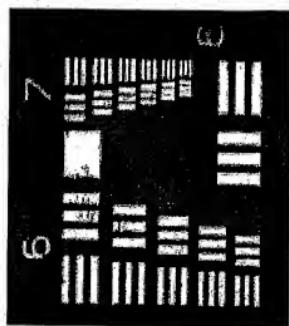


FIG. 5

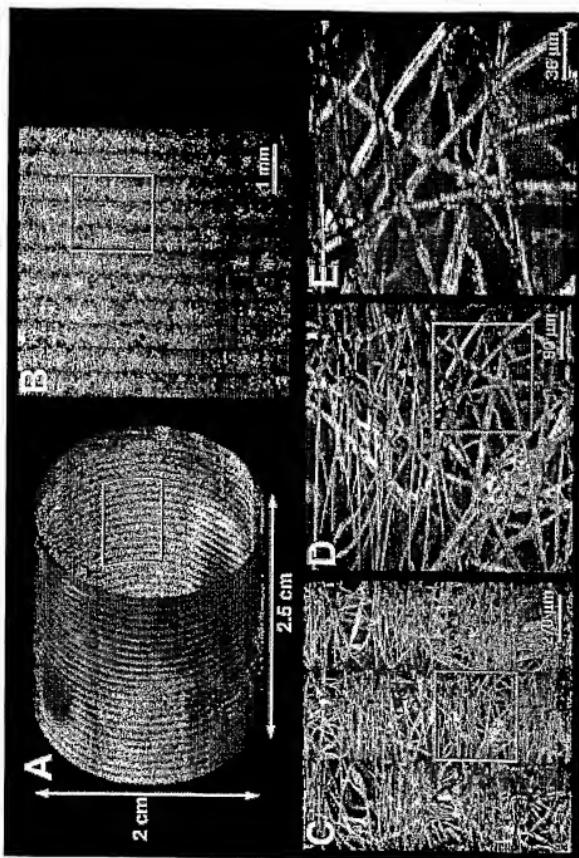


FIG. 6

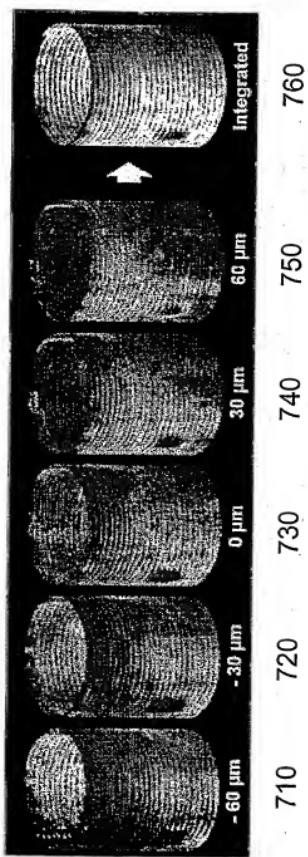


FIG. 7

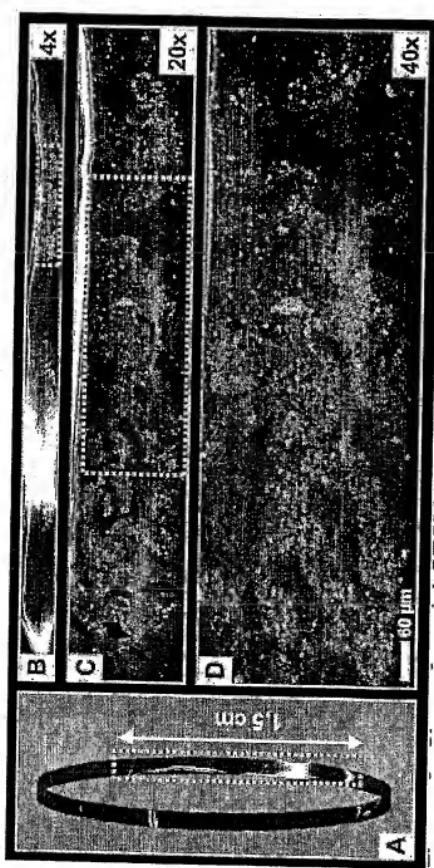


FIG. 8

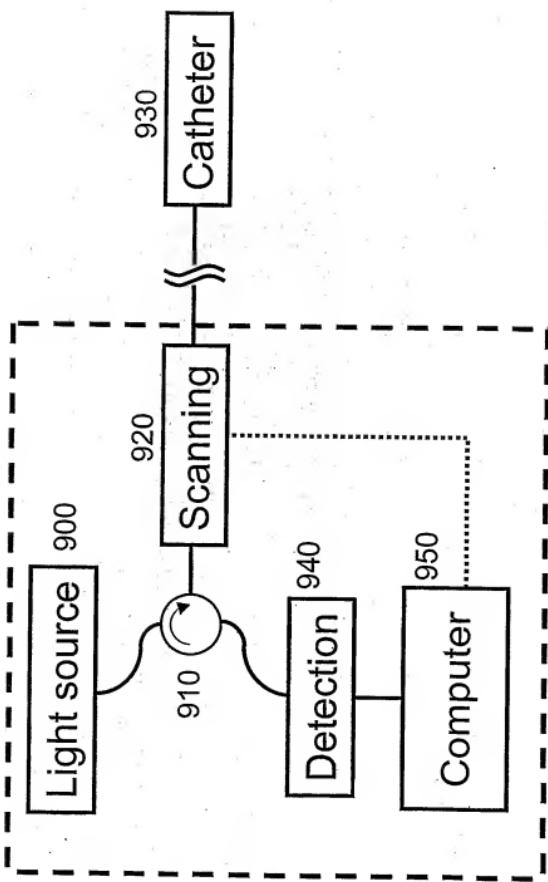


FIG. 9

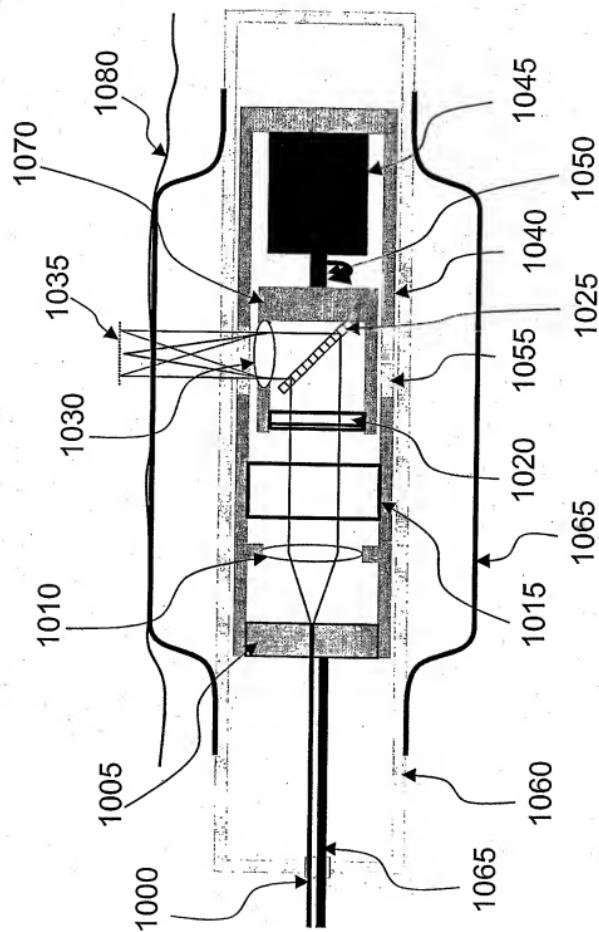


FIG. 10

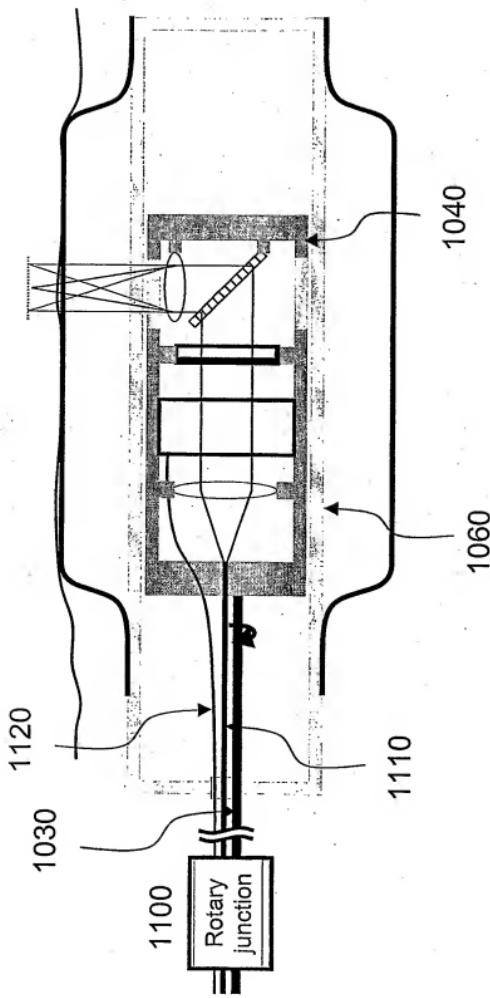


FIG. 11

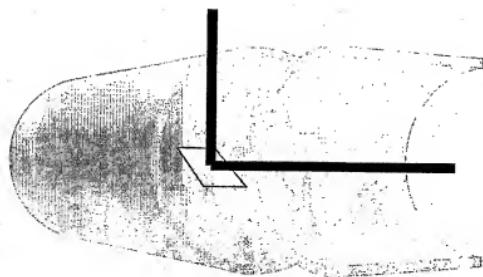


FIG. 12B

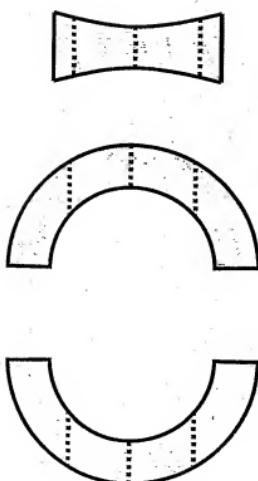


FIG. 12A

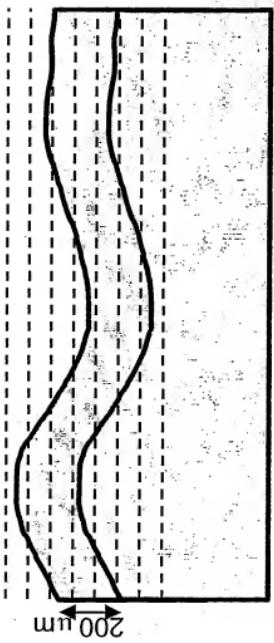


FIG. 13A

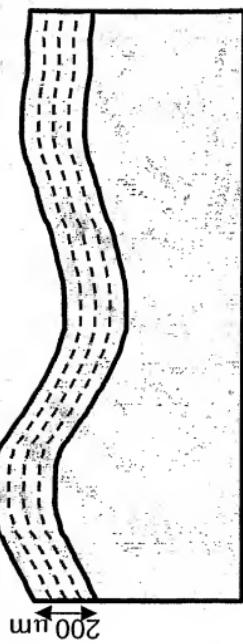


FIG. 13B

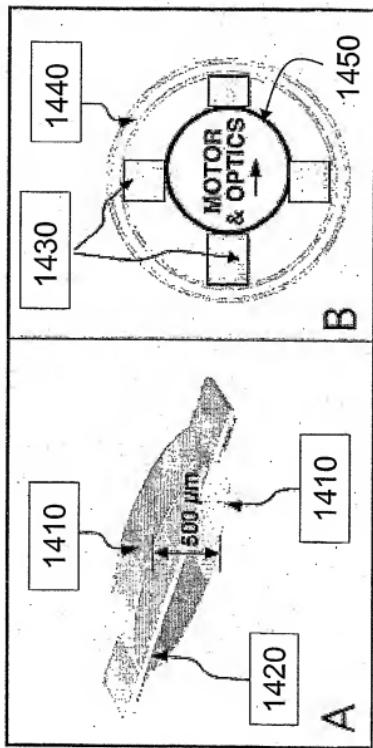


FIG. 14

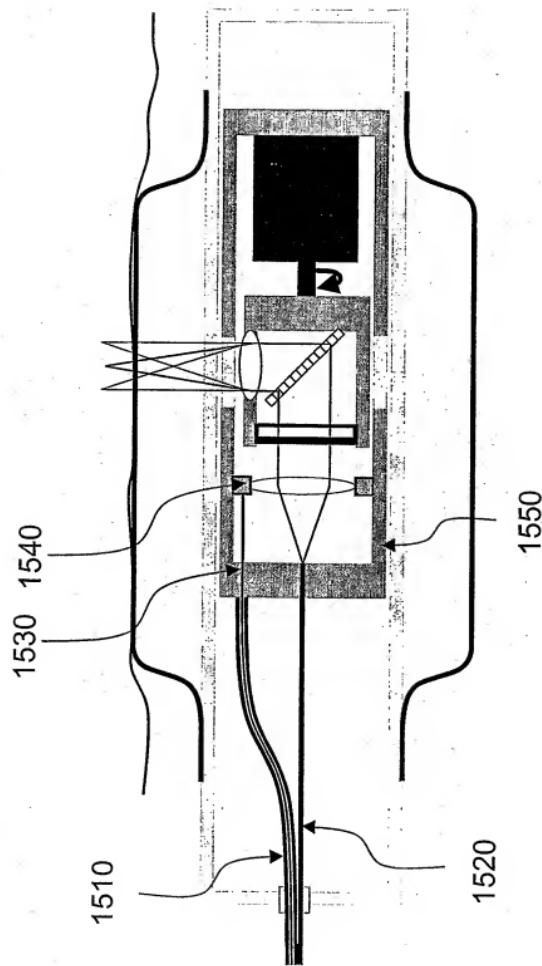


FIG. 15

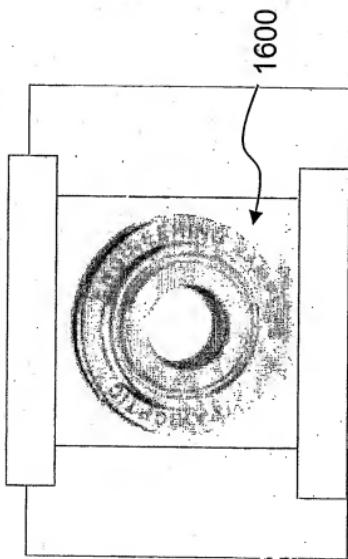


FIG. 16

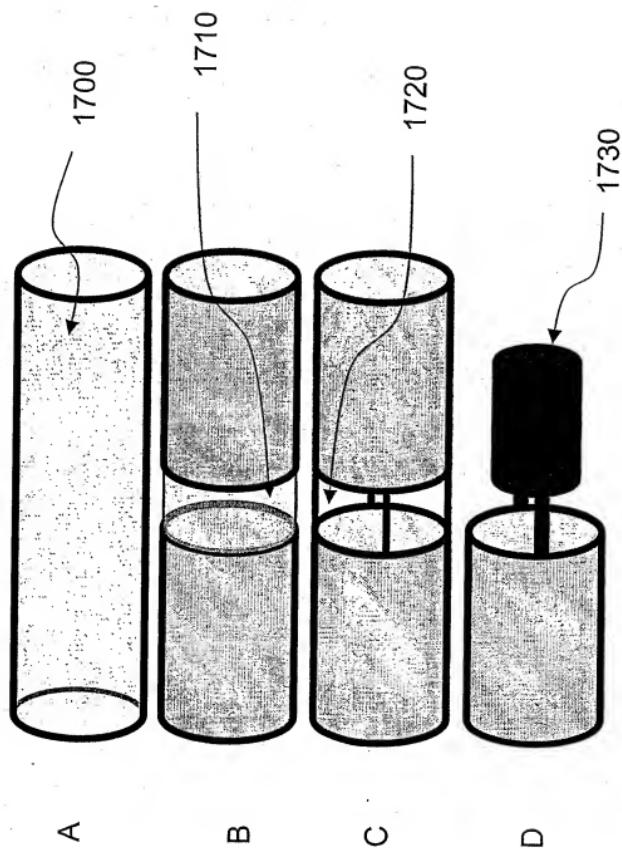


FIG. 17

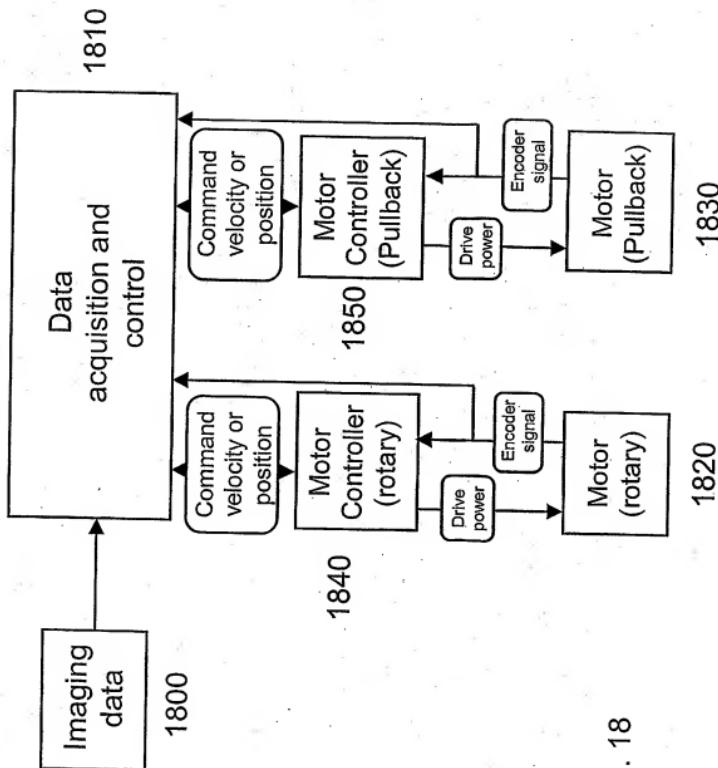


FIG. 18

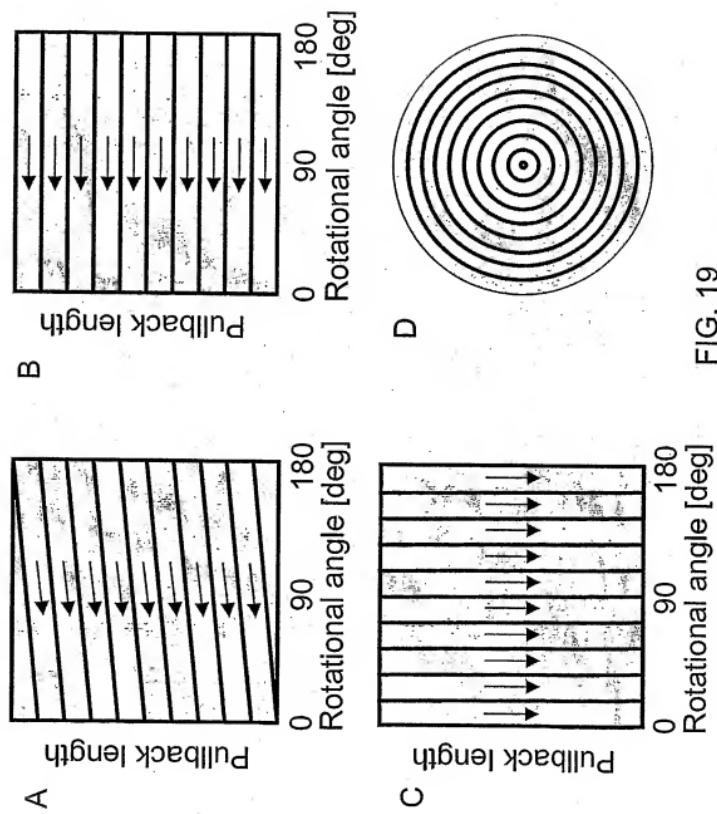


FIG. 19

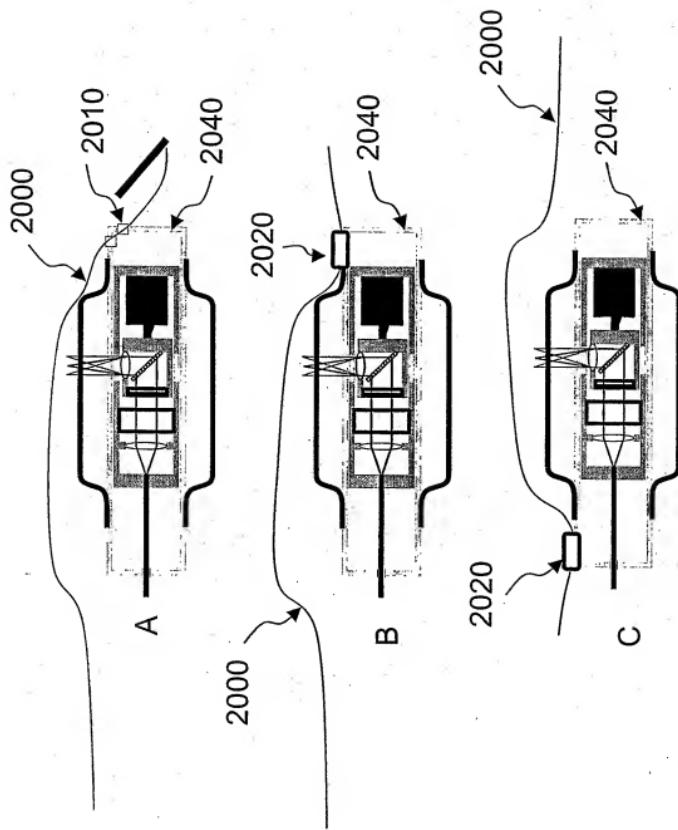


FIG. 20

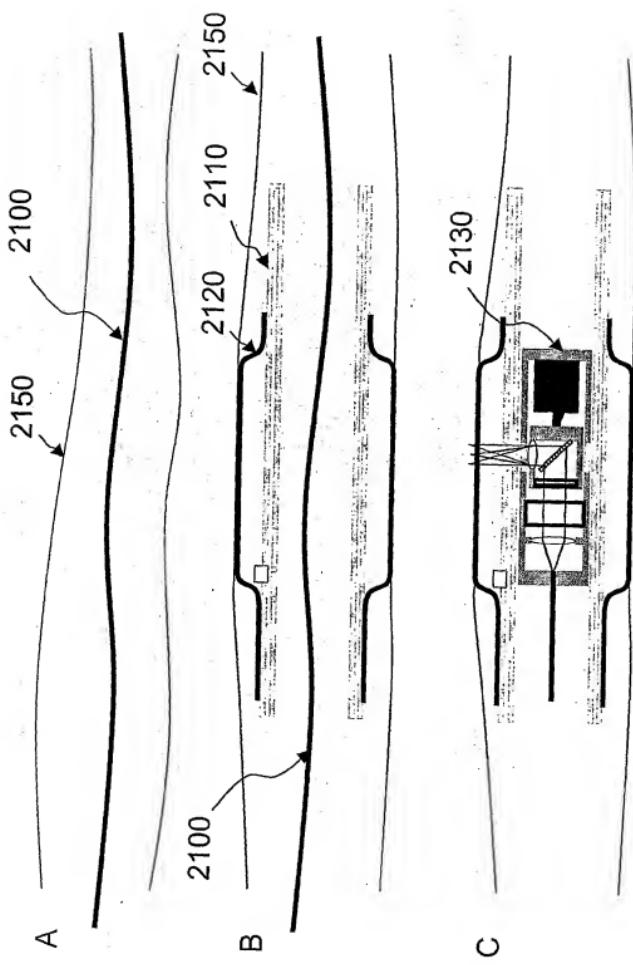


FIG. 21

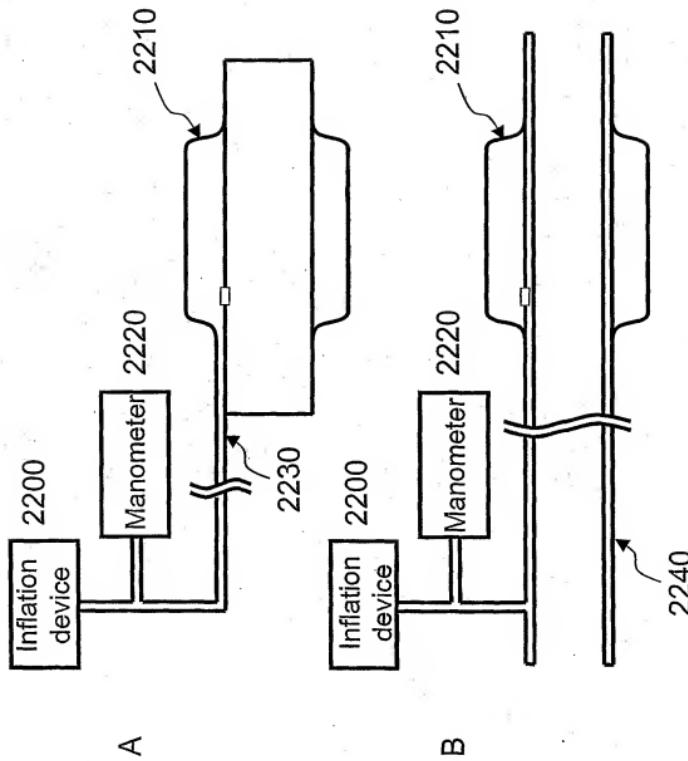


FIG. 22

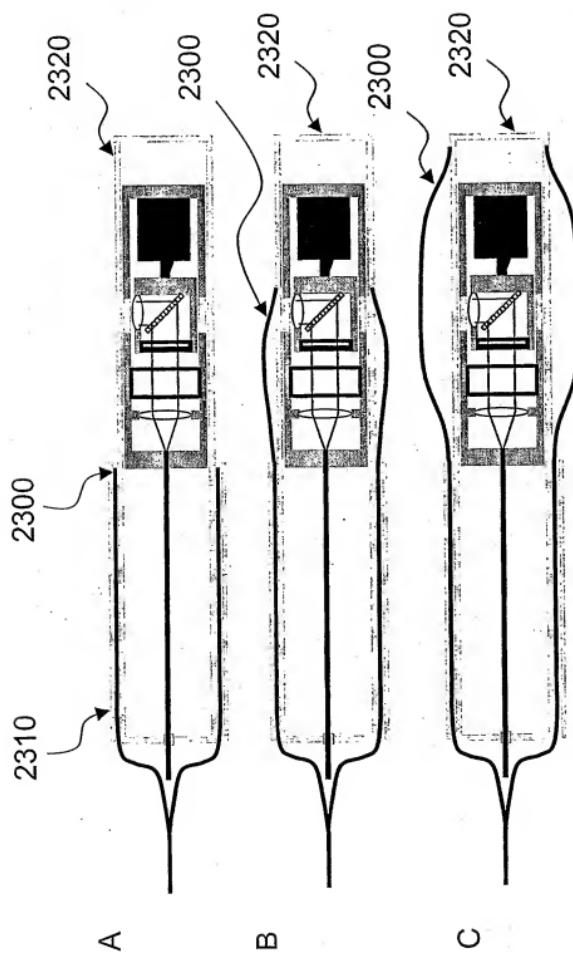


FIG. 23

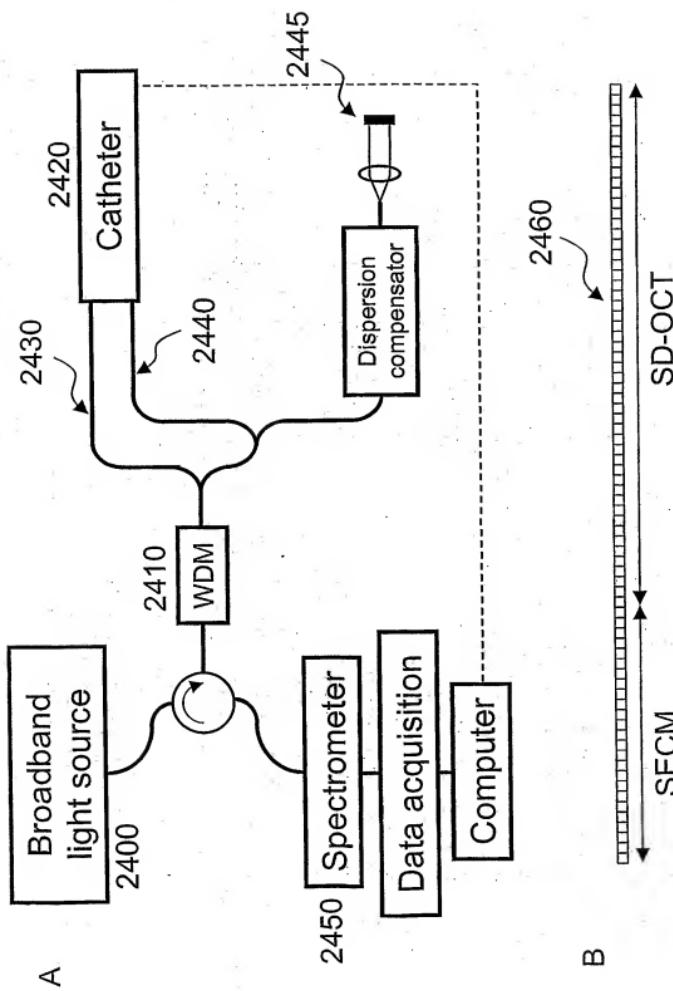


FIG. 24

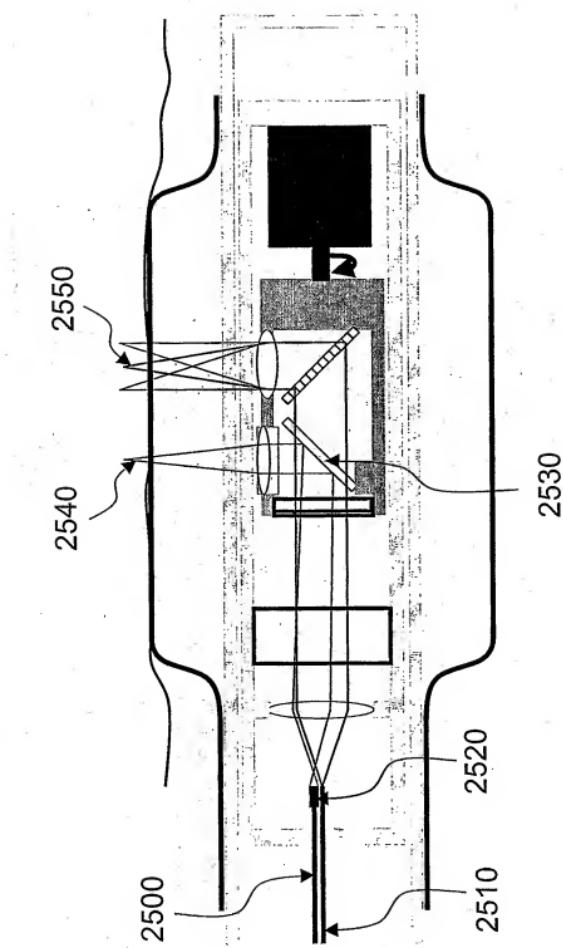


FIG. 25

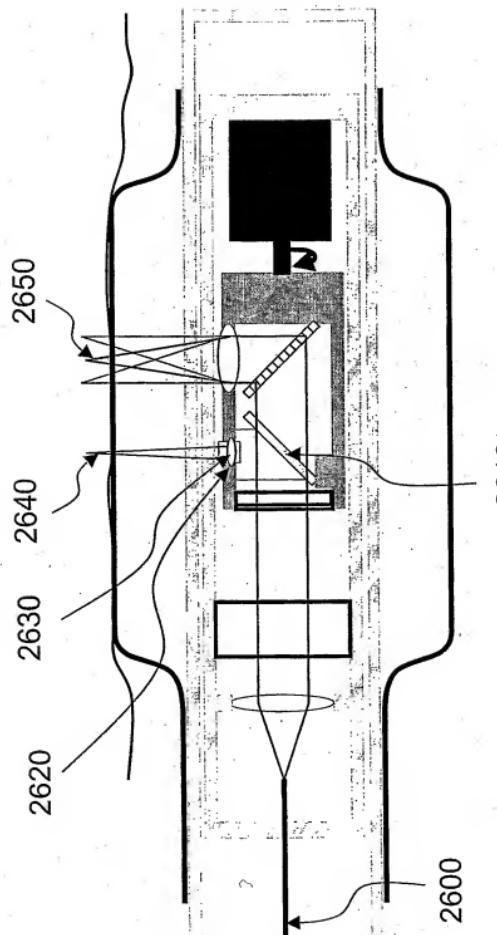


FIG. 26

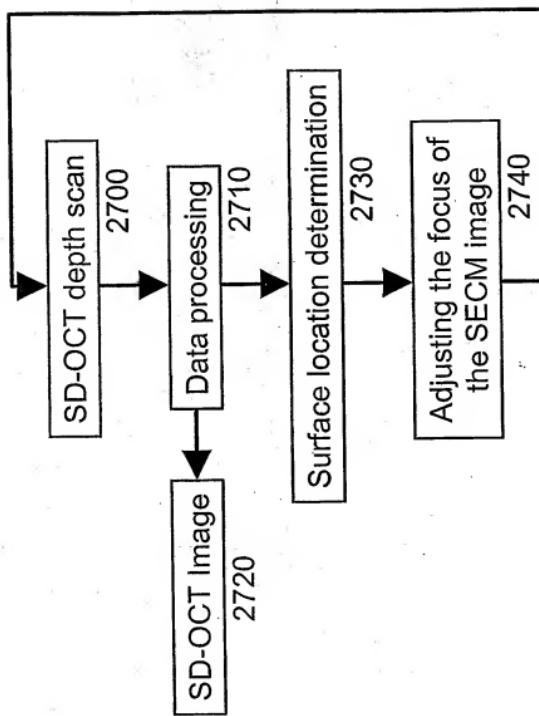


FIG. 27

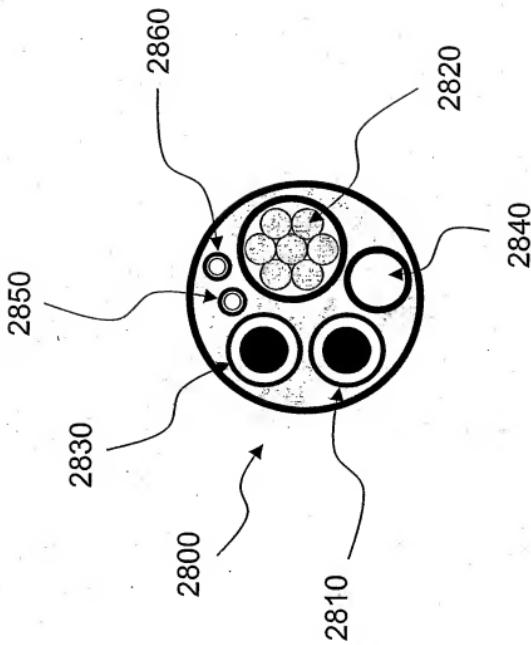


FIG. 28

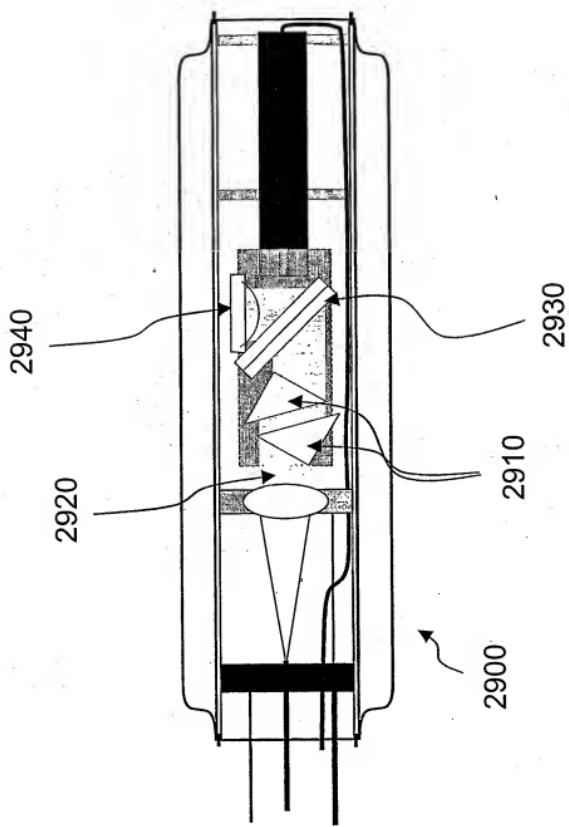
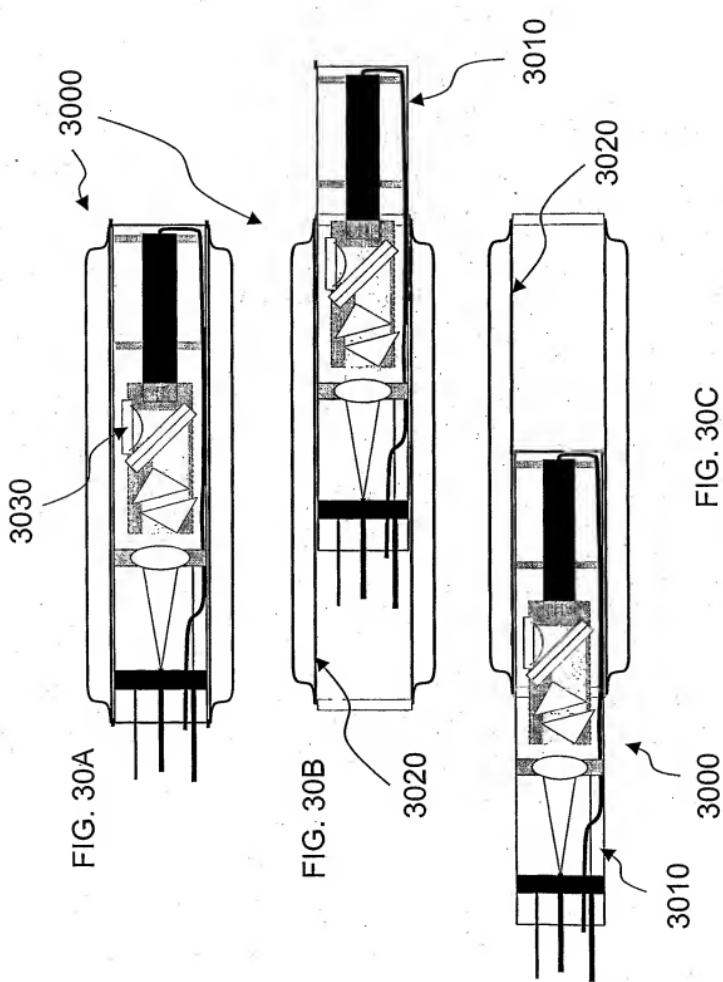


FIG. 29



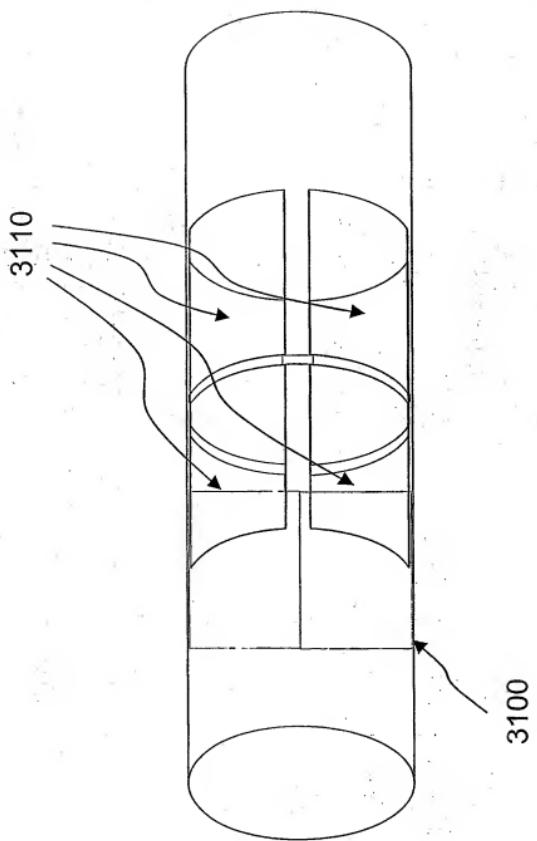


FIG. 31

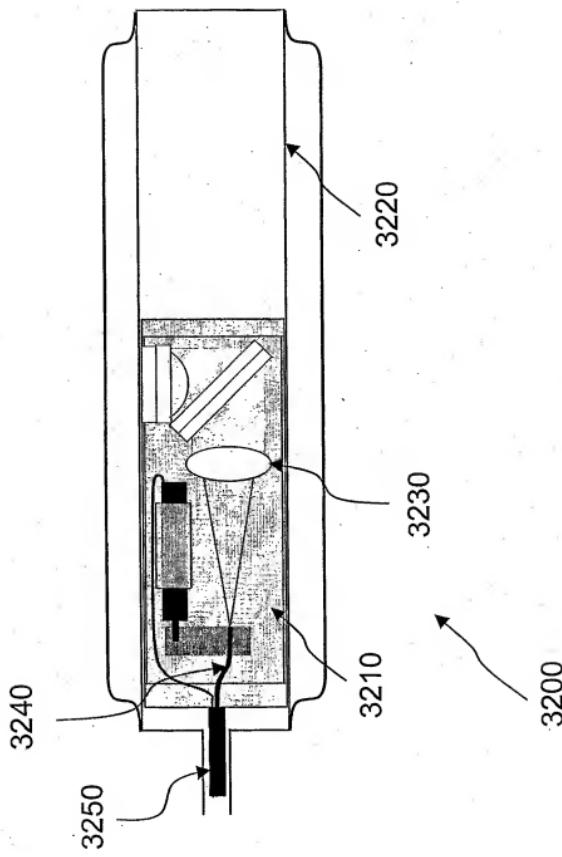
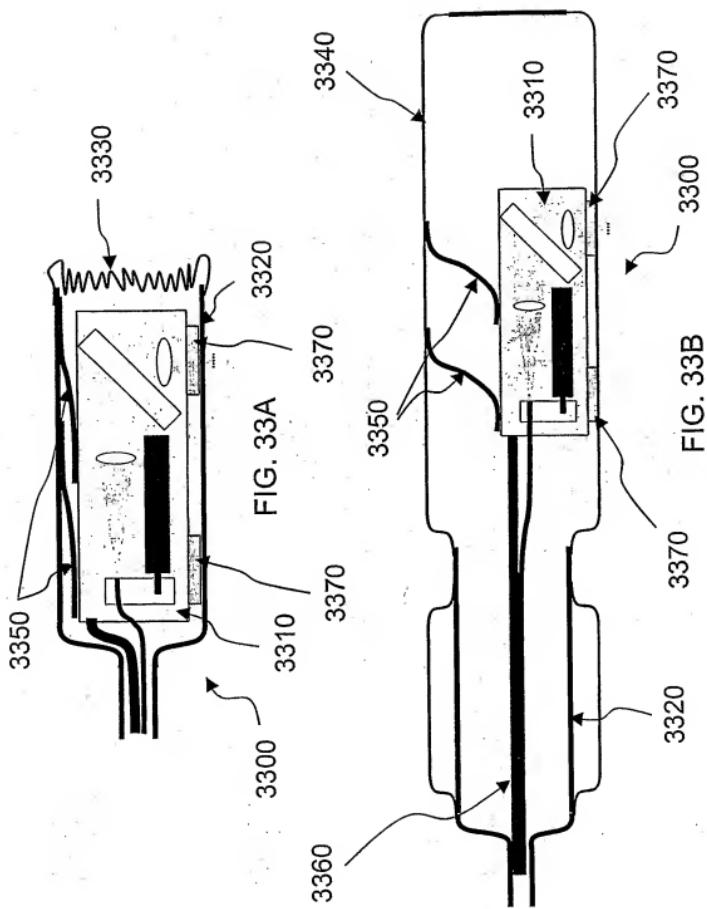
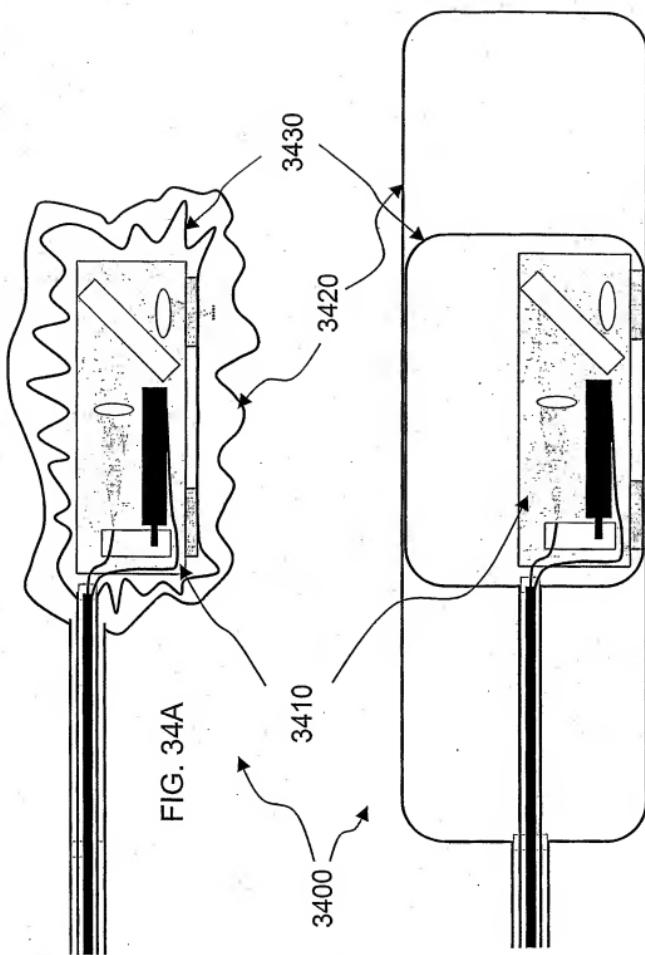
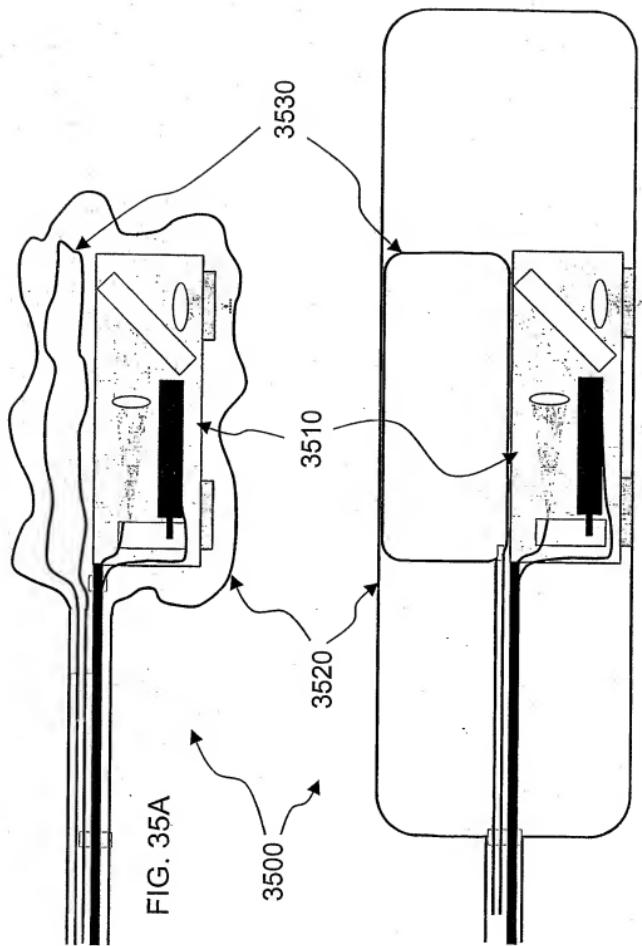
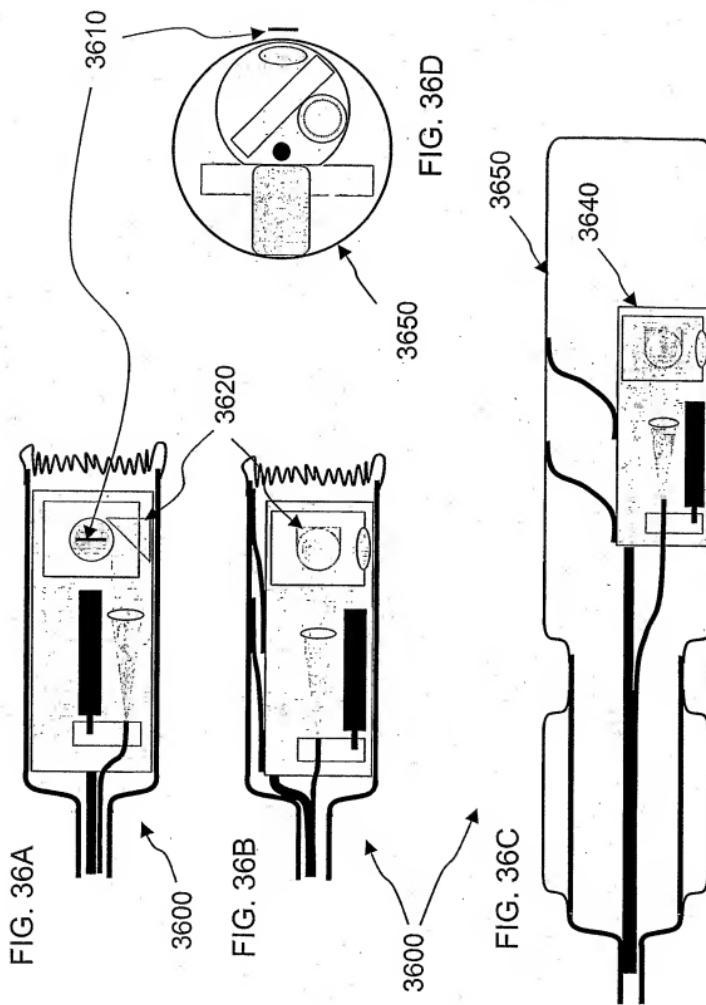


FIG. 32









A. CLASSIFICATION OF SUBJECT MATTER
 INV. A61B5/00 A61B1/07 GO1B9/02

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
 A61B GO1B GO1N

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

EPO-Internal, WPI Data

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	US 6 485 413 B1 (BOPPART STEPHEN A [US] ET AL) 26 November 2002 (2002-11-26) the whole document	1-43
X	US 2003/206321 A1 (GELIKONOV VALENTIN M [RU] ET AL GELIKONOV VALENTIN M [RU] ET AL) 6 November 2003 (2003-11-06) the whole document	1
X	WO 02/38040 A2 (GEN HOSPITAL CORP [US]; TEARNEY GUILLERMO J [US]; BOUMA BRETT EUGENE I) 16 May 2002 (2002-05-16) the whole document	1
X	US 5 715 825 A (CROWLEY ROBERT J [US]) 10 February 1998 (1998-02-10) figure 4	1
		-/-

Further documents are listed in the continuation of Box C.

See patent family annex.

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Date of the actual completion of the international search

13 February 2007

Date of mailing of the international search report

01/03/2007

Name and mailing address of the ISA/

European Patent Office, P.B. 5618 Patenttaan 2
 NL- 2280 HV Rijswijk
 Tel. (+31-70) 340-2040, Tx. 31 651 epo nl,
 Fax. (+31-70) 340-3016

Authorized officer

Plouzennec, Loïg

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	WO 99/44089 A (GEN HOSPITAL CORP [US]; WEBB ROBERT H [US]; TEARNEY GUILLERMO J [US];) 2 September 1999 (1999-09-02) the whole document	1-43

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			EP	1343411 A2		17-09-2003
US 5715825	A	10-02-1998	NONE			
WO 9944089	A	02-09-1999	AU	758078 B2		13-03-2003
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